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FOREWORD

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Blake H. Ford 26-May-99

PI - Signature Date

Table of Contents

Report Documentation Page

Foreword

1. Introduction	2
2. Research Accomplishments	3
2.1 Development of Surgical Instruments	3
2.2 Experimental Methodology – Databases of Force/Torque Signatures	3
2.3 Biomechanics and Statistical Models of the Data	4
2.4 Tissue Damage	5
3. Key Research Accomplishments	5
4. Reportable Outcomes	6
4.1 Manuscripts, Abstracts , Presentations	6
4.2 Database	7
4.3 Funding	7
4.4 Employment, Research and Training Opportunities	7
5. Conclusion	7
6. References	9
7. List of personnel	9
Appendices	10

1. Introduction

In traditional surgery the surgical tool has two interfaces: (i) surgeon/tool and (ii) tool/tissue. Assuming that the tool is a rigid body in quasi-static conditions (neglecting tool dynamics), the forces, torques and positions applied by the surgeon's hand on the surgical tool are transmitted directly to the tissue in contact with the tool. The reaction forces and torques are reflected back from the tissue/tool interface through the tool directly to the surgeon's hand. In this setup there are no additional forces that distort reliable flow of haptic and kinesthetic information in 6 degrees of freedom (DOF) from the tissue to the surgeon's hand through the tool.

In Minimally Invasive Surgery (MIS) a surgeon operates with specially designed surgical tools through access ports requiring incisions of about 1cm in size for placing the trocar. The trocar is a sealed tube inserted into the abdominal wall and used as a port for the endoscopic tool. The endoscopic tool, inserted into the body through the trocar, enables the surgeon to manipulate the internal organs by using the tool's finger loops outside the patient's body, which are linked to the tool tip via a long tube/shaft covering the tool's internal mechanisms. The trocar is a new interface in addition to the two that exist in traditional "open" surgery. Its inner tube surface interacts with the endoscopic tool through the sealing membrane, along with the outer surface interactions with the abdominal wall. The new boundary conditions generated by the trocar reduce the surgical tool's DOF from six (for open surgery) to four (for MIS). However, the constraint forces are not straightforward to compute since the abdominal wall is not rigid. Moreover, the trocar changes the way the surgeon interacts with the tissue through the tool. The action/reaction at the tool/hand interface is the sum of forces, torques, and angular displacements occurring at two locations: (i) tool tip/tissue of internal organ and (ii) trocar/tool shaft.

Surprisingly, little is known qualitatively about these forces and torques. Interest has recently increased because of the active development of virtual reality (VR) training aids for surgery. VR technology can be roughly divided into two subgroups: (i) VR where the user interacts with the virtual environment using only visual feedback (ii) VR with visual and force feedback in which the user can not only see the virtual objects but also feel the reaction forces when he/she comes into virtual contact with one. Such devices convey a kinesthetic sense of presence to the operator. The latter method incorporates a haptic device that generates the reaction forces. Since there is no experimental database of force/torque signatures in MIS. Current VR simulations are designed by guesswork and evaluated subjectively by an expert surgeon. The database measured in this research will allow the development of more objective methods for VR simulator evaluation.

The current project aimed to improve the scientific understanding of the biomechanics and human factors of surgery. Two interfaces out of the three that exist in minimally invasive surgery were studied: (i) endoscopic tool-tip / tissue and (ii) surgeon hand / endoscopic tool. The first task involved development of an endoscopic surgical grasper with computer control and a force feedback (haptic) user interface. Studying this two interfaces was accomplished

by two different type of measuring system: (i) Force feedback endoscopic grasper for measuring tool/tissue interaction and (ii) instrumented endoscopic grasper for measuring force/torque signature at the surgeon -hand/tool.

Experimental results demonstrated the ability to discriminate between normal tissue of small bowel, lung, spleen, liver, colon, and stomach. The second task used a six axis force-sensing laparoscopic grasper to measure forces and torques during a standardized surgical procedure. These measurements show distinct differences between expert and novice surgeons. Novices spent more time performing each maneuver and expended a greater magnitude of forces and torques to perform a task. We have also developed a software package, which allows for a simplified use of the computationally intensive Hidden Markov Model analysis. These results will have application to telesurgery, clinical endoscopic surgery, surgical training, and research.

2. Research Accomplishments

2.1 Development of Surgical Instruments [Ref 1,2,5]

During the course of this grant two different sensing surgical instruments were developed tested and used to develop a database.

The first instrument is a computerized Force Feedback Endoscopic surgical Grasper (FREG) having computer control and a haptic user interface. The FREG can control grasping forces either by surgeon teleoperation control, or under software control.

The second instrument is a regular endoscopic grasper which incorporated two sets of sensors for measuring the forces and torques at the interface between the surgeon's hand and the endoscopic grasper handle. The first sensor was a three axis force/torque sensor (ATI- Mini model) which was mounted into the outer tube (proximal end) of a standard reusable 10 mm endoscopic grasper. The sensor was capable of simultaneously measuring three components of force (F_x , F_y , F_z) and three components of torques (T_x , T_y , T_z) in the Cartesian frame. The second force sensor was mounted on the endoscopic grasper handle. Moving this handle caused the rod, sliding inside the outer tube, to transmit grasping/spreading forces from the surgeon's hand to the tool tip.

2.2 Experimental Methodology – Databases of Force/Torque Signatures [Ref 2,3,5,6]

The FREG performance was evaluated using an automated palpation function (programmed series of compressions) in which the grasper measured mechanical properties of the grasped materials. The material parameters obtained from measurements showed the ability of the FREG to discriminate between different types of normal soft tissues (small bowel, lung, spleen, liver, colon and stomach) and different kinds of artificial soft tissue replication

materials (latex/Silicone) for simulation purposes. The database includes force/deflection of internal organs grasped by a Babcock grasper tool tip.

The second database includes forces/torques applied by fourteen surgeons (seven novice surgeons and seven experienced surgeons) while performing laparoscopic cholecystectomy and laparoscopic Nissen fundoplication in a porcine model (pig). Each operation was divided into steps. Although all the steps were performed in each procedure, data were recorded only when the grasper was used with the following tool tips: atraumatic grasper, curved dissector, Babcock grasper (3 steps out of 4 in the first procedure and 3 out of six in the second procedure)

The seven channels of force/torque data were sampled at 30 Hz using a laptop with a PCMCIA A/D card. In addition, a LabView (National Instruments) application was developed incorporating a user interface for acquiring visualizing the force/torque data in real-time. The second source of information was the visual view from the endoscopic camera monitoring the movement of the grasper while interacting with the internal organs/tissues. This visual information was integrated with the force/torque human interface using a video mixer in a picture-in-picture mode and synchronized with time. The integrated interface was recorded during the surgical operation for off-line state analysis.

2.3 Biomechanics and Statistical Models of the Data [Ref 2, 5]

The viscoelastic theory of soft tissue developed by Fung was used to characterize the compression stress developed in the tissue as a function of the compression ratio. This relationship was expressed by an exponential equation with two parameters. Different parameter values were found using a curve fitting for characterizing the biomechanics of the internal organs soft tissues under compression conditions.

Two types of analysis were performed on the force/torque data measured at the surgeon's hand/tool interface: (I) video state analysis (SA) encoding the type of tool-tip/tissue interaction into states and (II) vector quantization (VQ) encoding the force/torque data into clusters (signatures). Each step of the operation was further divided into 17 different discrete tool maneuvers (states) in which the endoscopic tool was interacting with the tissue. Each identified surgical maneuver (state), had a unique force/torque pattern. For example, in the laparoscopic cholecystectomy, isolation of the cystic duct and artery involves performing repeated pushing and spreading maneuvers which in turn requires application of pushing forces mainly along the Z axis (F_z) and spreading forces (F_g) on the handle. Two expert surgeons independently performed frame by frame state analysis of the videotape with similar results.

The second type of analysis of the force/torque data used a VQ algorithm to encode the multi dimensional force/torque data (F_x , F_y , F_z , T_x , T_y , T_z , F_g) into discrete symbols representing clusters (signatures). First the 7D force/torque data vector was reduced to a 5D vector by calculating the magnitude of the force and torque in the XY plane (F_{xy} , T_{xy}). The K-means

algorithm was used to cluster the data into force-torque signatures of each one of the states defined in Table 1. Each force/torque signature represents a cluster center in a 5 dimensional space.

While we had hoped to establish statistical parameters that represented “optimal” applications of force and torque during an operation, we have found that we need more data to achieve that goal. We are still in the process of obtaining this data.

2.4 Tissue Damage

A controlled tissue damage was performed on different internal living pig organs: liver, spleen, serosa and muscular layer of stomach, small bowel, large bowel, and kidney. These organs were exposed to 3 levels of grasping forces (20,30 40 N) for 3 periods of time (30,60 90 sec) which applied by using a Babcock grasper in realistic surgical settings.

The specimens were then preserved in formalin and examined in a blinded fashion by a veterinary pathologist who was asked to grade the tissues. The results found were inconsistent and showed no logical progression of damage as determined by nuclear streaming, fibrin deposition, or acute hemorrhagic changes. We had postulated, prior to doing these experiments, that there would be consistent changes and that they would be further evidenced in chronic tissues which were exposed to an insult in a survival model with harvesting of tissues two weeks after the insult. Given the inconclusive results in our initial experiments, this line of inquiry was not further pursued.

3. Key Research Accomplishments

- Tools for measuring force/torque interaction between endoscopic tool and the tissue and between the tool and the surgeon hand. Database of stress/compression-ratio of internal organs.
 - Database of forces/torques signatures at the tool/hand interface generated in actual operating conditions of minimally invasive animal surgery in vivo.
 - Methodology for evaluation of surgery performance.
 - Biomechanical models of tissue under compression conditions.
 - Statistical state model of minimally invasive surgery.
-

4. Reportable Outcomes

4.1 Manuscripts, Abstracts , Presentations

Journal Papers

- Rosen J., B. Hannaford, M. MacFarlane, M. Sinanan, Force Controlled and Teleoperated Endoscopic Grasper for Minimally Invasive Surgery - Experimental Performance Evaluation, In press: IEEE Transactions on Biomedical Engineering, 1999.
- MacFarlane M., J. Rosen, B. Hannaford, C. Pellegrini, M. Sinanan, Biological and Simulated Soft Tissue Force Profiles Generated From a Force Feedback Grasper System, Submitted: Surgical Endoscopy.
- MacFarlane M., J. Rosen, B. Hannaford, C. Pellegrini, M. Sinanan, Force Feedback Grasper Helps Restore the Sense of Touch in Minimally Invasive Surgery, Accepted for Publication: Journal of Gastrointestinal Surgery.
- Richards C., J. Rosen, B. Hannaford, M. MacFarlane, C. Pellegrini, M. Sinanan, Skills Evaluation in Minimally Invasive Surgery Using Force/Torque Signatures, Submitted: Surgical Endoscopy.

Conference Papers

- Hannaford B., J. Trujillo, M. Sinanan, M. Moreyra, J. Rosen, J. Brown, R. Lueschke, M. MacFarlane, Computerized Endoscopic Surgical Grasper, Proceedings MMVR-98 (Medicine Meets Virtual Reality), San Diego, January 1998.
- Rosen J., M. MacFarlane, C. Richards, B. Hannaford, C. Pellegrini, M. Sinanan, "Surgeon/Endoscopic Tool Force-Torque Signatures In The Evaluation of Surgical Skills During Minimally Invasive Surgery, Proceedings MMVR-99 (Medicine Meets Virtual Reality), San Francisco, January 1999.

Presentations

- Hannaford B., J. Trujillo, M. Sinanan, M. Moreyra, J. Rosen, J. Brown, R. Lueschke, M. MacFarlane, Computerized Endoscopic Surgical Grasper, MMVR-98 (Medicine Meets Virtual Reality), San Diego, January 1998.
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- Rosen J., M. MacFarlane, C. Richards, B. Hannaford, C. Pellegrini, M. Sinanan, "Surgeon/Endoscopic Tool Force-Torque Signatures In The Evaluation of Surgical Skills During Minimally Invasive Surgery, MMVR-99 (Medicine Meets Virtual Reality), San Francisco, January 1999.
 - MacFarlane M., J. Rosen, B. Hannaford, C. Pellegrini, M. Sinanan, Biological and Simulated Soft Tissue Force Profiles Generated From a Force Feedback Grasper System, Meeting of the Society of American Gastro Intestinal Endoscopy, SAGES, Seattle WA, April 1998
 - MacFarlane M., J. Rosen, B. Hannaford, C. Pellegrini, M. Sinanan, Force Feedback Grasper Helps Restore the Sense of Touch in Minimally Invasive Surgery, Submitted: Journal of Gastrointestinal Surgery, Digestive Disease Week, SSAT Scientific, New Orleans LA, May 20 1998)

4.2 Databases

- Stress/compression-ratio of internal organs measured by minimally invasive grasper
- Forces/torques measured at the human-hand/endoscopic-tool interface generated by surgeons during selected steps of two minimally endoscopic procedures: laparoscopic Nissen fundoplication and laparoscopic cholecystectomy

4.3 Funding

Funding applied for based on this work

- NSF - "Abdominal wall/trocar interactions" – \$ 300k, 3 years
- NSF IGERT Program Preproposal – "Surgical science training program" – \$ 2.5M, 5 years

4.4 Employment, Research and Training Opportunities

Fellowships - Mark MacFarlane MD, Christina Richards MD

Post-Doc – Jacob Rosen Ph.D.

Training – Jeff Brown, undergraduate student

5. Conclusion

Part of the haptic information that is lost when a surgeon manipulates a soft tissue using an endoscopic tool/grasper may be regained by using the bilateral force feedback technology implemented in endoscopic instrument. The FREG is capable of controlling the force or

displacement of the jaw (tool tip) with interchangeable tools. To minimize cost and complexity, the system works with existing interchangeable reusable tools. The FREG controller was designed to maximize position control gain while preserving stability under unloaded conditions. Separating the human interface (finger loops) from the endoscopic tool allows one to generate a new Human-Machine interface (transfer function) in a way which enhances performance by overcoming the distortion which exists in the current mechanical endoscopic grasper setup.

The FREG in automatic mode is capable of discriminating between different soft tissues and latex/Silicone simulated tissue, and that the material's intrinsic biomechanical parameters can be identified for compression conditions. Moreover, a correlation from the mechanical characteristic perspective between the latex material and the soft tissue was found.

The statistical analysis of psychophysics performance aspects measured for the 10 test operators ranking 6 materials according to their stiffness suggests significant improvement in the performance of the FREG relative to a standard endoscopic grasper. The FREG performance was closer to the human hand, in rating materials stiffness, which defines the upper performance limit, than the standard endoscopic grasper which defines the lower limit. Even in the hand in glove conditions, the test operators were not capable of ranking the material stiffness correctly in all the cases. This fact may raise the need for more advanced instruments like the FREG capable of increasing the haptic sensation beyond the capability of unaided hand.

Minimally invasive surgery is a complex task which requires a synthesis between visual and haptic information. Analyzing MIS in terms of these two sources of information is a key step towards developing objective criteria for training surgeons and evaluating the performance of a master/slave robotic system for teleoperation or a haptic device for virtual reality simulations. The state transition data and the force/torque signatures are objective criteria for evaluating skills and performance in MIS. In general, it took the expert surgeon less time while applying less forces and torques to perform a typical MIS compared to the novice surgeon. This may be a result of advanced knowledge of the anatomy, higher level of eye-hand coordination and greater experience in handling the endoscopic surgical instrument.

The approach outlined in this study could be extended by increasing the size of the database which will allow development of statistical models like the Hidden Markov Model (HMM) of surgical procedures. This information, combined with other feedback data, may be used as a basis to develop teaching techniques for optimizing tool usage in MIS. The novice surgeons could practice these skills outside of the operating room on animal models or by using realistic virtual reality simulators, until they had achieved the desired level of competence, and compare themselves to norms established by experienced surgeons.

6. References

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- [5] Rosen J., M. MacFarlane, C. Richards, B. Hannaford, C. Pellegrini, M. Sinanan, "Surgeon/Endoscopic Tool Force-Torque Signatures In The Evaluation of Surgical Skills During Minimally Invasive Surgery, Proceedings MMVR-99 (Medicine Meets Virtual Reality), San Francisco, January 1999.
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7. List of personnel

List of personal received pay from the research effort

Prof. Blake Hannaford – PI

Prof. Mika Sinanan – PI

Mark MacFarlane MD – Fellow

Jacob Rosen Ph.D. – Research Associate

Christina Richards MD - Fellow

Jeff Brown – Undergraduate summer student

Appendices

Computerized Endoscopic Surgical Grasper

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Abstract. We report a computerized endoscopic surgical grasper with computer control and a force feedback (haptic) user interface. The system uses standard unmodified grasper shafts and tips. The device can control grasping forces either by direct surgeon control, via teleoperation, or under software control. In this paper, we test an automated palpation function in which the grasper measures mechanical properties of the grasped tissue by applying a programmed series of squeezes. Experimental results show the ability to discriminate between the normal tissues of small bowel, lung, spleen, liver, colon, and stomach. We anticipate applications in tele-surgery, clinical endoscopic surgery, surgical training, and research.

1. Introduction

As endoscopic procedures have rapidly grown in volume, surgeons have lost the ability to palpate tissues and organs. The corresponding diagnostic information is lost. Two components of this palpation information are tactile and kinesthetic information. Their combined use is referred to as haptic perception. Early work explored automated palpation with an external robot (1). Tactile sensors have been applied to endoscopic graspers which are coupled to tactile displays (2, 3). These systems aim to enable the surgeon to discriminate textural or time varying features of the patient via endoscopic tools. Morimoto et al., (4) have described an instrumented Babcock grasper which measured forces and torques at the tool-tissue interaction point, but did not measure or control grasping force.

The importance of haptic feedback to safe performance of surgery cannot be overstated. Although color, texture, and visible aspects of tissue deformation in the surgical field convey important anatomic information, palpation is critical to identifying otherwise obscure tissue planes, arterial pulsations, and regions of tissue thickening that may signify pathology such as infection or cancer. Safe tissue handling requires tissue manipulation that is both secure and nondamaging to the tissues. Much of the art of surgery and the implicit learning curve for traditional surgical technique depend on training to refine and educate the sense of

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touch. Training for endoscopic surgery is even more difficult because of the remote nature of videoendoscopic tissue manipulation. Indeed, recent literature emphasizes the importance of tactile feedback for accurate targeting of primary (5, 6, 7) and metastatic cancer (8, 9, 10, 11) and identifying therapeutic margins for curative resection (6, 12, 13). The loss of palpation for localization may seriously limit the efficacy and safety of minimally invasive treatment in some operative fields (11).

1.1. Purpose

This project aims to develop and characterize a grasper capable of restoring a degree of kinesthetic information to the surgeon about the tissue being grasped. The following goals were laid out:

1) Improve the ability of the endoscopic surgeon to feel mechanical properties of tissues such as compliance.

2) Make minimal changes to the form and function of existing surgical graspers to reduce cost, complexity, and certification difficulties. Avoid adding sensors and wiring to the tool tip.

3) Take advantage of the declining cost of computer control.

Our system is designed to support both manual and automatic palpation. For reasons of space, this report will describe mainly the automatic function.

2. Methods

The present grasper (Fig. 1) is a re-design of the handle end of an existing stainless steel, reusable, interchangeable grasper. The tool head consists of the tool shaft mount, electro-

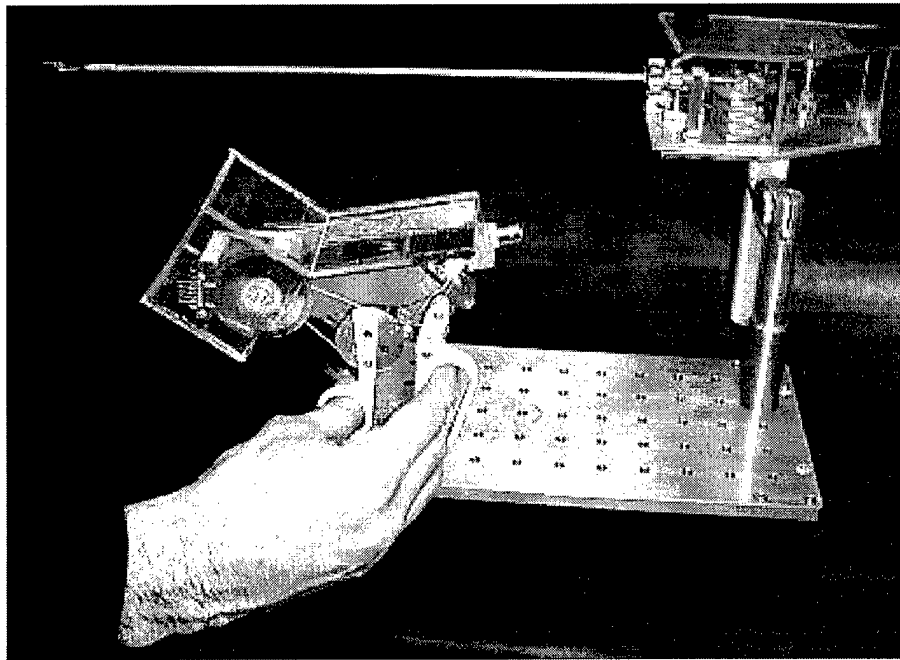


Figure 1. Photo of the current design of the computer controlled endoscopic grasper mounted on its storage base.

magnetic actuator (see below), and an optical encoder position transducer. These elements are mounted on a handle for the surgeon which can be attached and detached from a base. Also on the base is a separate user interface consisting of finger loops taken from another grasper. The distal finger loop is connected to an actuator/encoder pair identical to those on the tool shaft. To increase sensing resolution, the encoder wheels are connected to the actuation axes via pulleys and a kevlar drive belt having a multiplication ratio of 1:3.6. As a consequence, both master and slave have 1400 quadrature position counts over the full 0.6 radian (34.4 degrees) motion range.

The actuators are flat coil actuators modified from hard disk drive head positioning actuators. In an earlier prototype, the actuators were taken directly from 5.25 inch (133mm) hard drives. Hard disk drive head actuators have many advantages for precision robotics and force feedback devices (15). However in this application, the actuators' maximum torque of 0.1NM at 2.0 amps (based on steady state coil temperature of 93 deg. C) did not produce convincing subjective grasping sensations. The actuator magnets were replaced with custom made Nd-Fe-B magnets having approximately triple the energy product of the AlNiCo magnets used in the disk drive actuator (14). The coil and bearing assembly was retained. To realize the full flux increase from the new magnets, we built new frames from high permeability iron to prevent backing iron saturation. The new actuator magnets and frames increased the torque output to 0.3 NM but preserved the desirable qualities of low torque ripple, low friction, and low backdriving inertia.

The laparoscopic instrument used in these experiments is a stainless steel atraumatic Babcock grasper (Carl Storz Inc., model # 30420 BL) with a square jaw grasping surface area measuring 9 x 9 mm. The tool shaft is 5 mm in diameter and 38 cm long from the proximal attachment to the instrument tip. The shaft and mount allow 360 degree rotation of the tool about its long axis. The proximal end of the instrument shaft is clamped to a supporting post on the slave handle. The push rod operating the jaws is linked to the electromagnetic actuator via a ball and socket joint. This system allows easy change of shaft length, diameter, and tool tip conformations. Laparoscopic tools compatible with the mounting system are readily available from various manufacturers.

3. Control

The control system supports both bi-lateral force reflecting teleoperation of the grasper jaws, and programmed automatic operation for tissue characterization. Proportional-derivative (PD) controllers were designed for both the master and slave using a linear dynamic model of the device and conventional control techniques (16). Integral feedback is not desirable in position error based force feedback control because it creates a time varying force feedback under conditions of steady state contact.

The force feedback controller is based on the well known bi-lateral, position error based, teleoperation system (17). In this design, the measured position of each side serves as the reference position input for the other.

A desirable quality of force feedback systems is a high effective stiffness between master and slave sides. In the position error based architecture, this requirement can be translated into the need for a high value of the proportional feedback gain, K_p (18). An additional controller design constraint is introduced from the actuator limit of 0.3NM maximum torque. Experience shows that users feel a subjective loss of contact sensations when

a force feedback device saturates at its maximum force output. There is thus a trade-off between K_p and the deflection at which saturation occurs. For high values of K_p , the user will feel high effective stiffness, but saturation will occur at relatively smaller position errors. We set the position error corresponding to the saturation point at one quarter of the motion travel range which in turn sets

$$K_p = \frac{I_{max}}{0.16rad} = 12.6 \quad (1)$$

The remaining parameter K_d was determined by placing the dominant closed loop pole for an 8 ms settling time constant and a damping ratio of 0.5. For the slave, this design method resulted in an unstable controller, possibly because of backlash in its mechanism. An acceptable controller was recomputed with a lower initial K_p value. The resulting gains are given in Table 1.

Table 1: Controller Parameters

	$K_p \left(\frac{NM}{rad} \right)$	$K_d \left(\frac{NMsec}{rad} \right)$
Master	12.6	0.05
Slave	9.6	0.04

3.1. Automatic Palpation Mode

Because the grasper is computer controlled, the possibility exists to create automated grasping and palpation functions in software. This could be used for automating surgical functions such as grasping with a pre-set force level, or for quantitative, automated palpation in which the deflection and force measurements are analyzed to extract information about tissue mechanical properties. Our initial experiments were designed to evaluate the information which can be obtained by driving the slave position controller with a sinusoidal displacement command while recording position, position error, and torque command. Other testing modes will be evaluated in future work such as applying a torque command and recording displacement. In the experiments reported here, three cycles of a 1 Hz sinusoidal displacement were applied as the desired position input to the slave controller. The amplitude of the sinusoid corresponded to full opening and closing of the jaws (0.6 rad).

Full analysis of the solid mechanics of the Babcock grasper interacting with organ tissue is beyond the scope of this paper, but compared to other types of surgical grasping instruments, the geometry of the Babcock tool suggests that it creates a relatively uniform stress distribution under the contact sites. In a future report (19) we will describe the analysis method in more detail.

Torque vs. displacement data were first isolated in time to the segment involving initial contact and compressive displacement. Next, considering the grasper mechanism, and a modified version of the theory proposed by Fung (20) for viscoelastic material, the torque-displacement data measured at the handle were transformed to the uniaxial compression stress-length ratio. Then, the stress-length ratio data were fitted, using the Least-Square method, with Eqn. 2.

$$\sigma = -\beta (e^{\alpha(1-\lambda)} - 1) \quad (2)$$

$$\sigma = \frac{F}{A} \quad (3)$$

$$\lambda = \frac{L}{L_0} \quad (4)$$

where: α and β are parameters, λ is the compression length ratio, σ the uniaxial compression stress [Pa], A the compression cross section area [m^2], L the Length of the material compressed by the load [m]. L_0 is the length of the material at zero load [m], and F is the compression force applied by grasper tip [N]. The resulting parameters α and β are features of the tissue as computed from the graspers measurements. Generally speaking, higher values of α and β describe "stiffer" tissues.

Protocols for anesthetic management, euthanasia, and survival procedures were reviewed and approved by the Animal Care Committee of the University of Washington and the Animal Use Review Division of the U. S Army Veterinary Corps.

In addition to pig tissues, five different latex materials were examined. For the purpose of further discussion they were designated MAT1, MAT2, MAT3, MAT4, MAT5. All the latex materials were shaped in the same cylindrical form with a diameter of 13 mm and a length of 45 mm. The above designation was referred to each material by a subjective estimation of its stiffness where MAT1 is the softest material and MAT5 is the stiffest material. MAT1 to MAT4 can be considered viscoelastic materials representing artificial replication of soft tissues while MAT5 can be defined as a solid which exhibits the upper limit of physiological stiffness and can simulate the bone tissue.

4. Results

To analyze the data recorded by the automatic palpation function, the stress-length ratio curves for the compression phase of each material squeeze (tissues and latex) were fit with the exponential function (Eqn. 2) using the least squares method.

Most of the recorded data were well fit by (Eqn. 2). The quality of the numerical fit was verified by using the correlation ratio factor, R^2 . The computed R^2 values were typically very close to one ($R^2 > 0.999$), indicating very high quality of fit between (Eqn. 2) and the experimental data. Two exceptions with relatively lower R^2 values ($R^2 > 0.99$) were the colon tissue and the stomach. Those tissues exhibited different type of compression characteristics especially at lower compression length ratios.

Since the software generated three squeeze/open cycles, there were three squeezes recorded 1 second apart for each grasp. Tissues typically got stiffer in the second and third squeezes of each sequence.

Scatter plots were made of the α and β parameters for the pig tissue and latex material (Fig 2). Data formed into clusters. Each cluster consists of nine data points. Rectangles defined by the univariate standard deviations computed from the organ data clusters did not overlap except for lung and spleen. These variances are partly due to the stiffening of tissues under repeated compression as described above.

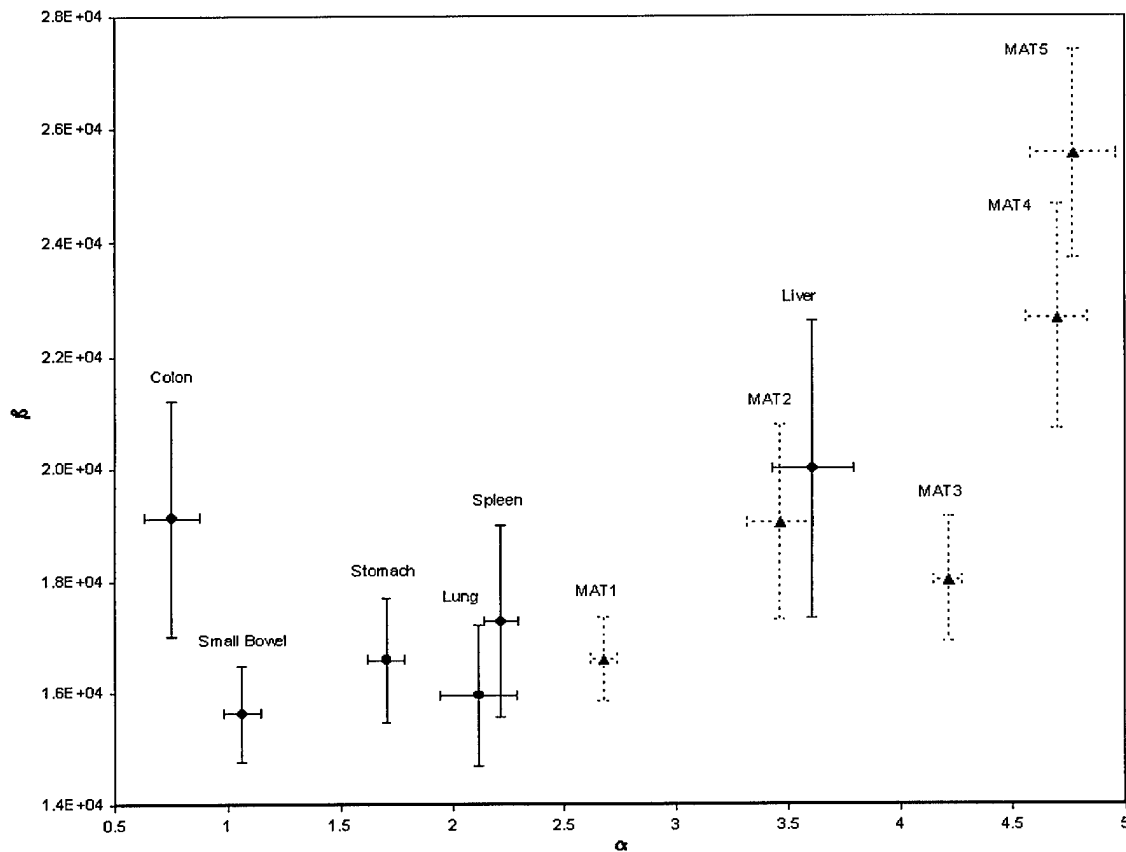


Figure 2. Scatter plot of the curve fitting parameters (Alpha, Beta) for different pig tissues (solid) and latex materials (dashed).

5. Conclusions and Discussion

We have reported a modified surgical grasper capable of controlling the force or displacement of jaw opening with interchangeable tools. To minimize cost and complexity, the system works with existing interchangeable re-usable tools. The controller was designed to maximize position control gain while preserving stability under unloaded conditions.

Initial tests revealed promising performance in the ability to reliably distinguish different tissue mechanical properties in the automated mode. In the future, we plan to measure its ability to distinguish pathological tissue from normal. An additional benefit to this project is an improved ability to characterize artificial materials for use in disposable organ simulators for surgical training. For example, Figure 2 indicates that "MAT2" might be a suitable material for simulating liver.

In additional preliminary experiments, beyond the scope of this paper, surgeons were able to distinguish the same tissues using the instrument in the teleoperated mode. A future study will quantify this ability in detail.

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Biological and simulated soft tissue force profiles generated from a force feedback grasper system

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Abstract

Training in laparoscopic surgery outside of the operating room setting is currently limited by the high cost and ethical considerations of animal models; lack of realistic simulated tissue models and limited force feedback in virtual reality models. We have developed a computerized endoscopic grasper system with force feedback capabilities which is able to generate tissue force profiles. Six different tissue types from live anesthetized pigs, and five latex material phantoms were analyzed by three computer-controlled sinusoidal compressions. These tests resulted in a calculated stress compression ratio for each biological and simulated tissue type. Scatterplots of these ratios showed minimal overlap within each biological or latex tissue subset. Comparison of biological to latex phantoms did show some close overlaps between biological and latex tissues. Based on this data, the stress compression ratios allow discrimination of tissues types. Force profiles give objective data for creating a surgical simulation environment to closely match the real characteristics of biological tissues in a virtual reality model. In addition, creation of latex materials which match the force profiles of biological tissues will also enhance realism in a simulated tissue model. We believe that these surgical models will play a large part in training future surgeons in open and endoscopic skills.

Key words: Force profile; Surgical simulation; Force feedback

Introduction

The current age of minimally invasive surgery has brought forth astounding changes in the health care system. Patients have benefited by faster recovery [7,9,20,23], less patient discomfort [11,15], improved cosmesis via smaller incisions and fewer complications [21]. Insurance companies and employers alike have also benefited by way of shorter hospital length of stay [4,8,12] resulting in lower hospital charges [6,14] and a quicker return to work [19].

In order to keep pace with this new technology, fully trained general surgeons had to gain new additional skills in order to remain competitive in the current rapidly changing health care system. These endoscopic skills have predominantly been obtained through a mentoring relationship with a well-trained endoscopic surgeon or through animal courses. But these two methods are usually time-limited because of the busy schedules of surgeons and the cost and ethical issues related to utilization of animal models. Some simulated models are available in attempt to develop endoscopic skills, but these models also lack the realism of endoscopic surgery, thus their effectiveness in producing endoscopic skills is in question [13,22]. More importantly, if surgeons are performing endoscopic procedures without adequate training because of the economic pressures to offer these procedures, then unnecessary mistakes and injury to the patient will undoubtedly occur [3,16,17,18]. To compound the problem, no uniformly recognized standard of training or level of skill has been set before a surgeon performs endoscopic surgery on patients.

In order to address the underlying problem of gaining endoscopic skills, we believe that an accurate and realistic simulation model is the safest and best method for initial training in endoscopic surgery. This would allow repetition of procedures until an adequate level of skill is obtained before proceeding to operations on humans. To create such a realistic simulation is complicated, but one aspect which we have attempted to address is to obtain the crucial information of haptics (the sense of touch), which will allow a simulated tissue to feel like a biological tissue when touched or grasped with an instrument. Through the use of haptic technology, we were able to measure the intrinsic force properties of biological tissues, and thus have the capability to create simulated tissues which can closely match biological force properties. This information is paramount in creating a realistic surgical simulation.

Materials and methods

Computerized grasper design

The design of the grasper involves a master and slave component which is linked by a computer interface. The master component is manipulated by the surgeon through a standard set of endoscopic finger loops. By this movement, the surgeon determines the corresponding position of the instrument tool tip on the slave component. The driving force to move the tool tip is a electromagnetic coil actuator on the slave. The ability of the master and slave components to determine the position of the tool tip and finger loops is carried out by means of identical optical encoder position sensors on the master and slave. Thus, as the surgeon manipulates the finger loops on the master, the position is measured by the master position sensor. This position is then transferred via the computer interface to the slave position sensor, and the slave actuator then moves the tool tip to the corresponding position of the finger loops. The force feedback capability (or sense of touch) of this device is produced by a second and identical actuator on the master which is linked to the finger loops. Therefore, as the slave actuator creates a force to move the tool tip and compress whatever is in the tool tip, the identical force is generated in the master actuator which is linked to the surgeon's hand by means of the finger loops. This constant and simultaneous interplay of position and force between the master and slave is outlined in figure 1.

The computer interface allows transmission of signals of position and force between the master and slave. In addition, it also allows recording of the force and position data, and thus allows one to measure how much force is required to displace a tissue a given distance. The computer interface also has the capability to perform a task with the slave in a automated preprogrammed manner (no involvement of master) or in a bimanual mode (response of the slave is controlled by the action of the master). For the purposes of standardization and reproducibility, all experiments reported here were carried out in a computer automated mode.

The tool tip used throughout the experiments consisted of an endoscopic babcock grasper tip and shaft with a tool tip surface area of 9x9 mm (Carl Storz Inc., model # 30420 BL). Although the design allows for easy interchange of standard endoscopic instrument types, the babcock grasper was used because it allowed a uniform method of compressing nearly all tissues sampled. The initial prototype is displayed in figures 2 and 3.

Biological tissue force profiles

Protocols for anesthetic management and euthanasia were reviewed and approved by the Animal Care Committee of the University of Washington and the Animal Use Review Division of the U.S. Army Veterinary Corps. Three month old pigs were anesthetized with isoflurane after endotracheal intubation. A laparotomy was then performed to allow exposure to the abdominal organs, later followed by a transabdominal incision of the diaphragm to allow access to lung tissue. In the automated computerized mode, each

tissue type was palpated by three consecutive one second duration sinusoidal cycles. This was repeated three times on each tissue for a total of nine force/displacement data points for each tissue. The six tissue types examined included colon, small bowel, stomach, lung, liver and spleen. The animals were given a lethal injection of pentobarbital at the completion of the experiments, thus pain and suffering to the animals was minimized.

Simulated tissue force profiles

Five latex samples of uniform shape (13mm diameter cylinders) and varying compliances were custom manufactured (Simulab Inc., Seattle, Washington) to simulate soft tissues. Identical palpation experiments were performed as outlined above in the animal experiments to generate force/displacement data points.

Data analysis

Force/displacement data was applied to the exponential equation: $\sigma = -\beta(e^{\alpha(1-\lambda)} - 1)$ where α and β are parameters, λ is the compression length ratio, and σ is the uniaxial compression stress (Pa). Scatterplots were generated from the alpha and beta parameters. This equation is based on work by Fung et al which describes the properties of soft tissue biomechanics [10].

Results

Data points for the six different biological tissues generated from palpation of the tissues were fitted into stress/compression curves as illustrated in figure 4. Increasing stress is applied to the tissues (y-axis) in order to produce a given compression of each tissue type. All six tissues have two curves. One curve being the ideal fit to Fung's equation, and the second curve being the actual fit of our data. Each tissue type fit correlates highly ($R^2 > 0.999$) with the ideal fit, thus our data is consistent with Fung's model of soft tissue biomechanics.

The same data was then fitted to a scatterplot format with 9 data points for each tissue type. This is displayed in figure 5. Clearly evident is the separation of tissue types based on Fung's model of soft tissue biomechanics. Furthermore, hollow organs group on the left side of the x-axis, while solid organs are present on the right side of the x-axis in the direction of increasing stiffness.

Scatterplots from palpation of the 5 different latex phantoms were generated and are displayed in figure 6. Only minimal overlap is seen between samples labeled L4 and L5. Of note is that L1 is the softest material and L5 is the stiffest material by subjective palpation.

Figure 7 displays the combined scatterplots of the biological organs and the latex phantoms (figure 5 and 6 combined). Of note, is that the latex tissue labeled L2 closely approximates the force profile of liver tissue.

Discussion

Endoscopic based operations continue to expand into new areas of surgery. Operations which were once thought possible to complete only open, are now being performed commonly in a endoscopic approach [5]. In order to continue performing endoscopic operations safely, optimal training techniques need to be available to surgical residents and seasoned surgeons who lack the specialized training of advanced endoscopic procedures. Although animal models are realistic and closely approximate human anatomy, their high cost, limited use, and ethical considerations do not make them an ideal model. The historical model of apprenticeship training, learning on humans, is probably not the best method either because of the consequences of mistakes.

Based on experience in other simulated training environments, a successful training model must be realistic, cost effective, and have the ability to be used repetitively until the trainee has mastered the desired skill. A virtual model has the potential of reaching such a goal, but cost performance and reliability issues need to be overcome before this is achieved [1,2]. One major area which we have chosen to investigate in order to improve simulation is the area of haptics, or the sense of touch or feel. Most designers of surgical simulation models have not collected haptic data to incorporate into their models, but have instead relied on subjective data to determine how organs are supposed to feel. Unfortunately, this method is inaccurate, and may create unrealistic and potentially dangerous force expectations if a trainee applies excessive force on a human organ based on inaccurate simulated training.

Although a discussion of haptic technology and soft tissue biomechanics is beyond the scope of this paper, with application of this technology we were able to produce force/displacement curves for the six different biological tissues tested with minimal overlap of data points (figure 4). In addition, the two curves for each tissue type (ideal fit and actual data fit) had a high correlation value $R^2 > 0.999$ in the setting of Fung's model for soft tissue biomechanics. Thus, we feel that our data is accurate and reproducible. Other variables outside of our control would expect to alter the force/displacement curves, such as tissue edema, metastatic disease, cirrhosis of the liver, scar formation, etc. This may be evident in figure 5, where force profiles of pig organs are displayed. Hollow organs conglomerate on the left side of the X-axis while solid organs group on the right side (increasing stiffness). Although hollow organs have greater macroscopic variability (mucosa, muscular layers, and serosa) than solid organs, they also contain displaceable luminal air which could explain their softer appearance. These additional variables are currently under investigation and are important in the study of diseased organs and the fragility of such altered organs.

In addition to the creation of virtual simulation models for surgical training, reusable/disposable latex models have also been developed to aid in surgical training. It would seem that latex models also need to look and feel realistic to be of maximum benefit. Figure 6 depicts force profiles on 5 different latex phantoms manufactured by a local company specializing in simulated soft tissues (Simulab Inc., Seattle, WA.). The clear separation of the phantoms based on their compliance is evident. Figure 7 shows an overlay of biological and phantom tissues. Even though overlap of force profiles between biological and latex phantoms only occurred between liver and the latex phantom labeled

L2, the manufacture of these latex products can be easily manipulated. For example, to create a latex simulated stomach, perhaps a hollow structure with a luminal air/latex interface would be created. Creation of latex phantoms to closely match the force profiles of the respective biological tissue should be easily achievable.

In summary, stress-compression ratios appear to characterize tissues uniquely. Realistic virtual and latex simulated models may to be created for training of surgical residents or seasoned surgeons without advanced endoscopic surgical skills. Haptic data from real tissues is essential in meeting this goal of maximum realism. With this data, we have established such a method that allows the accurate collection of haptic information for biological tissues to meet part of the needs for accurate simulation. Likewise, this model allows gathering of haptic information on altered or diseased tissues which can also be applied to surgical simulation models. The ultimate goal is to train better surgeons in an environment which is safer for patients.

Acknowledgments

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Legends

Figure 1. Flow diagram for interaction of master and slave components of the grasper system.

Figure 2. Master and slave components of the computerized grasper. Both parts are detachable from the metal base as needed.

Figure 3. Detailed views of the master and slave components. Each component contains an actuator and position sensor.

Figure 4. Stress compression ratios for each of six biological tissue types. Two curves for each organ are presented: one is the ideal fit, and the second is the actual data fit to Fung's model.

Figure 5. Force profile scatterplots of six biological tissues. Arrow indicates direction of increasing tissue stiffness.

Figure 6. Force profile scatterplots of five latex phantoms. Arrow indicates direction of increasing tissue stiffness.

Figure 7. Combined force profiles of latex phantoms and biological tissues. Bold lines indicate latex materials, and fine lines correspond to biological tissues.

Force-Feedback Grasper Helps Restore Sense of Touch in Minimally Invasive Surgery

Mark MacFarlane, M.D., Jacob Rosen, Ph.D., Blake Hannaford, Ph.D., Carlos Pellegrini, M.D., Mika Sinanan, M.D., Ph.D.

The age of minimally invasive surgery has brought forth astounding changes in the health care field. Less pain and quicker patient recovery have been demonstrated with several types of operations that were once performed by an open technique. With these changes have come reports of complications. The decreased sense of touch is just one of several limitations inherent to current techniques of minimally invasive surgery that limit detection of subtle or unapparent lesions on palpation, such as common duct stones and liver lesions. The purpose of this study is to demonstrate the ability of a force-feedback-equipped grasper to restore some of the sense of touch that is lost in minimally invasive surgery. To demonstrate this ability, we created six silicone phantoms of identical dimensions but graded compliance, and asked 10 subjects to place them in increasing/decreasing order of compliance. They used three tools (their dominant gloved hand, a standard laparoscopic Babcock grasper, and our force-feedback device fitted with the identical Babcock grasper) to rate the compliance of the samples in a blinded fashion. These conditions thus approximated the conditions of open surgery, minimally invasive surgery, and minimally invasive surgery fitted with a force-sensing device, in terms of palpating tissues. Five surgeons skilled in minimally invasive surgery and five nonsurgeons participated in the study. The results indicate that the force-feedback device is significantly ($P < 0.05$) better than a standard Babcock grasper at rating tissue compliance, but was not as successful as a gloved hand (mean of squared errors = 1.06, 3.15, and 0.25, respectively). There was no significant difference between surgeons and nonsurgeons in rating compliance. We conclude that this force-feedback instrument is able to partially restore the sense of touch in minimally invasive surgery. This restored ability may thus potentially result in more efficient operations with improved diagnostic capabilities and fewer complications during minimally invasive surgery. (J GASTROINTEST SURG 1999;3:278-285.)

KEY WORDS: Haptic, surgical simulation, force feedback, touch

The current age of minimally invasive surgery has brought forth astounding changes in the health care system. Patients have benefited by faster recovery,¹⁻⁴ less patient discomfort,^{5,6} and improved cosmesis because of the smaller incisions. Insurance companies and employers alike have also benefited by way of shorter hospital stays⁷⁻⁹ resulting in lower hospital charges^{10,11} and a quicker return to work.¹² Unfortunately this new technology has also been accompanied by reports of endoscopic complications such as gastrointestinal and colon perforation, as well as injuries to other organs.¹³ In addition, some diagnostic information may be lost when endoscopic surgery is performed¹⁴ because of the inability of the surgeon to feel

the tissues with the hand. This may result in underestimated or unrecognized tissue inflammation or inability to detect solid and hollow organ masses. The preceding disadvantages of minimally invasive surgery are a result (at least in part) of the need to use long instruments that leave the surgeon at a mechanical disadvantage in terms of the haptic interface or sense of touch. Other investigators have proposed ways to improve the haptic feedback in minimally invasive surgery by incorporating a sleeve,¹⁴ which allows passage of the hand into the abdomen, but this requires a larger incision, thus partly defeating the purpose of minimally invasive surgery. Visual cues can supply information on tissue deformation and com-

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pliance,^{15,16} but the information is highly subjective and incomplete.

To address the deficit of haptic feedback during minimally invasive surgery, we have designed and tested an endoscopic grasper with force-feedback capabilities,¹⁷ to improve the sense of touch in minimally invasive surgery. This prototype instrument is our initial attempt to enhance the sense of touch during minimally invasive surgery.

MATERIAL AND METHODS

Computerized Grasper Design

The grasper design utilizes "master" and "slave" components, which are linked by a computer interface. The master component is manipulated by the surgeon through a standard set of endoscopic finger loops. By this movement the surgeon determines the corresponding position of the instrument tool tip on the slave component. The driving force to move the tool tip is an electromagnetic coil actuator on the slave. The position of the tool tip and finger loops is measured by identical optical encoder position sensors on the master and slave. Thus, as the surgeon manipulates the finger loops on the master, the position is measured by the master position sensor. This position is then transferred via the computer interface to the slave position sensor, and the slave actuator then moves the tool tip to the corresponding position of the finger loops. The force-feedback capability (haptic interface or sense of touch) of this device is produced by a second and identical actuator on the master which is linked to the finger loops. Therefore as the slave actuator creates a force to move the tool

tip and compress whatever is in the tool tip, the identical force is generated in the master actuator, which is linked to the surgeon's hand by means of the finger loops. This constant and simultaneous interplay of position and force between the master and slave is outlined in Fig. 1.

The computer interface allows transmission of signals of position and force between the master and slave. In addition, it also allows real-time display and recording of force and position data, and thus allows one to measure how much force is required to displace a tissue a given distance. The computer interface also has the capability to perform a task with the slave in an automated preprogrammed manner (no involvement of master) or in a bimanual mode (response of the slave is controlled by the action of the master). The force-feedback mode is the bimanual mode. More technical aspects of the grasper system have been described in detail previously.¹⁷

The tool tip used throughout the experiments consisted of a nondisposable endoscopic Babcock grasper tip and shaft with a tool tip surface area of 9×9 mm. The prototype is displayed in Figs. 2 and 3.

Creation and Objective Testing of Silicone Phantoms

Six silicone phantoms of uniform shape and color (15 mm diameter \times 150 mm length cylinders) but of varying compliance (Fig. 4) were custom manufactured by a local private company (Simulab Inc., Seattle, Wash.). The compliance of the materials was altered by varying the percentage by weight of the catalyst during manufacture of the phantoms. To ob-

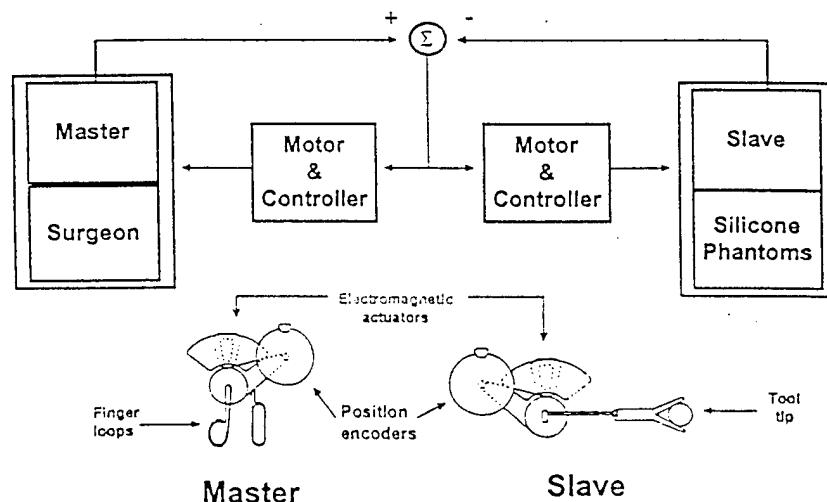


Fig. 1. Flow diagram for interaction of master and slave components of the grasper system. Optical encoder position sensors and electromagnetic actuators are identical on the master and slave components.

Fig. 2. Master and slave components of the computerized grasper. Both parts can be detached from the metal base as needed.

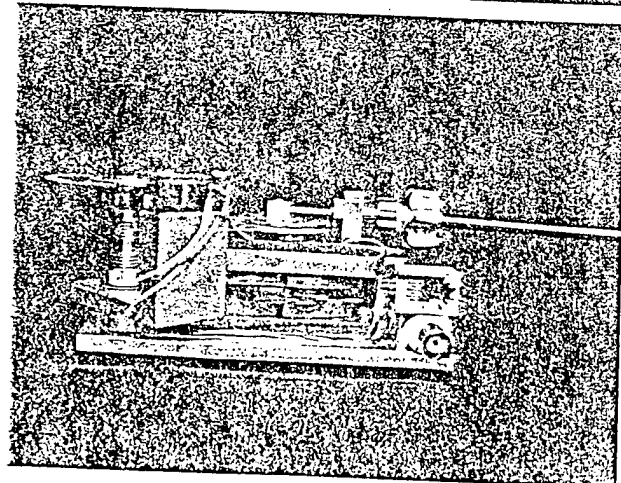
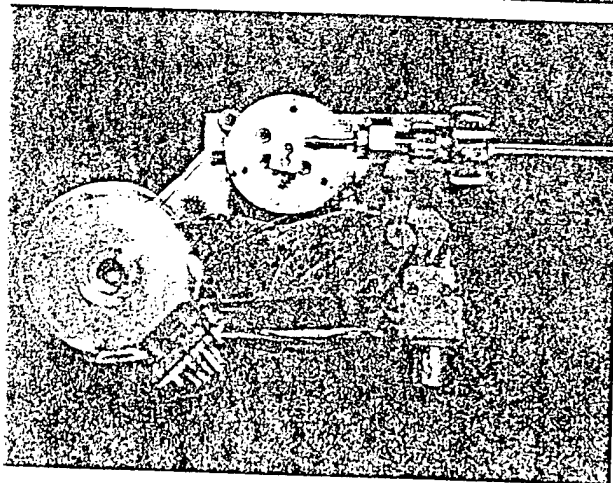
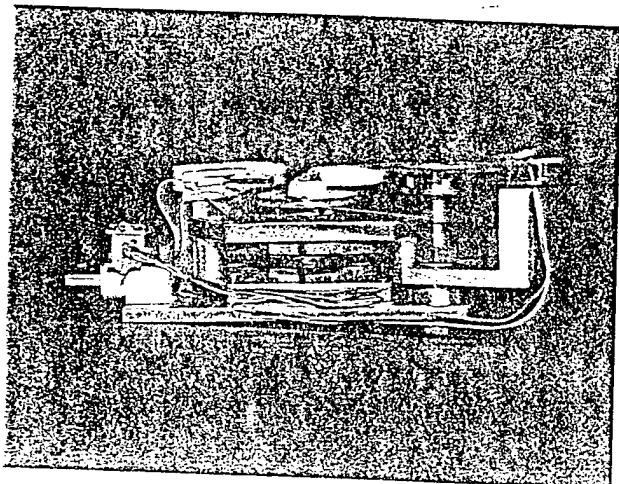
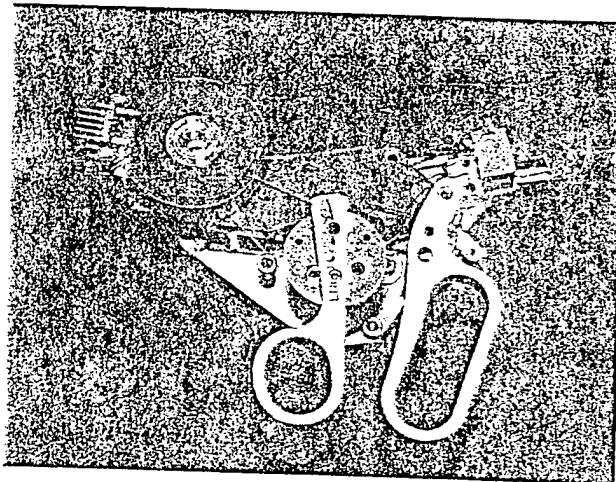
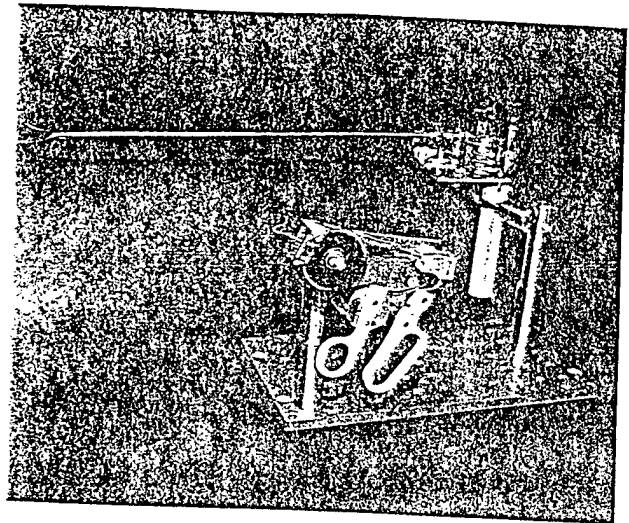


Fig. 3. Detailed views of the master (A and B) and slave (C and D) components. Each component can be detached from the metal base as needed.

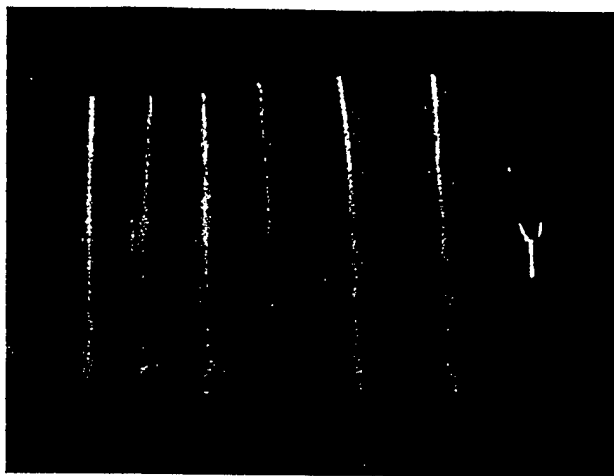


Fig. 4. Photograph of the six silicone phantoms of graded compliance. Dimensions are identical for each of the six phantoms.

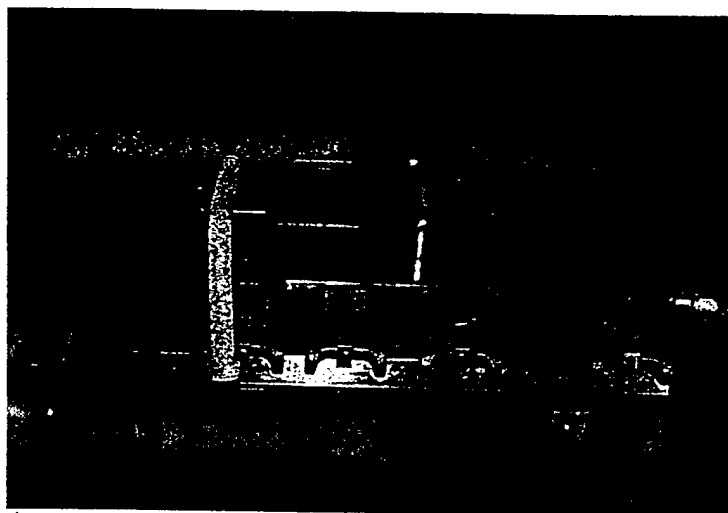


Fig. 5. Device used to carry out objective stress/compression studies of the six silicone phantoms.

tain an objective measure of the compliance ratings, the phantoms were subjected to stress/compression testing. This was performed by sequentially compressing each phantom in increments of 1.1 mm over a total distance of 9.9 mm using a device that held and compressed the materials. We measured the force required to compress the materials by means of a force meter attached to the compression device (Fig. 5). Testing was performed three times on each silicone phantom. The data were used to construct stress/compression curves for each phantom (Fig. 6).

Subjective Compliance Rating of Silicone Phantoms

To test for possible improvement in haptic feedback by the force-feedback grasper compared to a

standard laparoscopic instrument, we performed experiments as outlined below. Ten subjects (five experienced laparoscopic surgeons and five electrical engineers with experience in haptic technology) performed palpation experiments on the six silicone phantoms. Subjects were asked to place the phantoms in the correct order of increasing or decreasing compliance. This was performed a total of four times for each palpation tool used. The three tools for palpation included a dominant surgical gloved hand (simulating open surgery palpation), a standard 10 mm nondisposable laparoscopic Babcock grasper (simulating endoscopic surgical palpation), and an identical Babcock grasper tip fitted to our force-feedback device in the bimanual mode (simulating possibly improved endoscopic surgical palpation). Subjects were allowed a 3-minute unblinded practice period for each

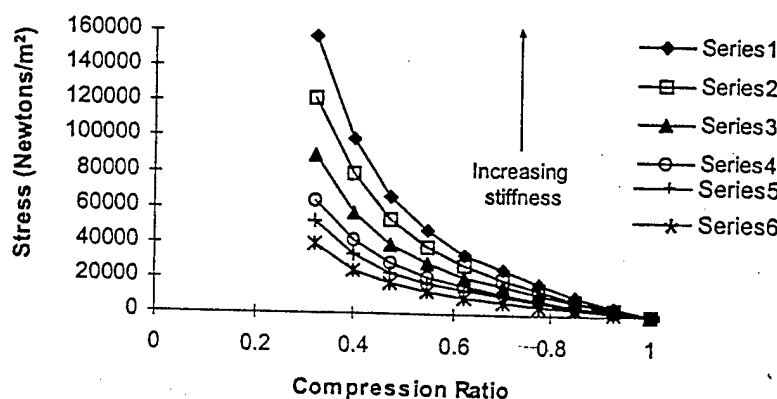


Fig. 6. Stress/compression ratios for each of the six silicone phantoms. Standard deviations (not shown) indicate minimal to no overlap, even at the low end of the compression ratio, although this is not appreciated on the graph secondary to the large y-scale.

of the three tools so they could become acquainted with the palpation of the phantoms. The phantoms were presented to the subjects in random order, and the order was changed on each presentation. The phantoms were assigned a number of 1 through 6 (1 = hardest, 6 = softest) based on the preceding objective compliance testing. Subjects attempted to choose the correct order. The difference of the chosen order (subjective) from the known order (objective) was then squared to give a positive integer for each choice. Each squared value was summed, and the total divided by six (6 phantoms) to yield the mean squared error. Testing resulted in a total of 12 rounds of compliance ratings for each of the six phantoms for each subject (a possibility of 72 errors in compliance rating for each subject and 720 possible errors for the 10 subjects). Although the tool type obviously could not be blinded, the subjects were not allowed to visualize any interaction of the tool type/phantom interface. Likewise the subjects were not allowed to visualize any part of the three tools or phantoms during the experiments; thus visual cues were eliminated in their subjective evaluation of phantom compliance.

Data Analysis

Objective stress/compression ratio curves were generated from 36 data points for each silicone phantom after graduated compression. A best curve fit using Matlab (The MathWorks, Inc., Natick, Mass.) was used to construct the stress/compression curves.

Subjective compliance ratings of the six silicone phantoms by 10 subjects were scored as the mean of the squared difference (error) of the subjective order

from the known correct order as described previously. Results were analyzed with a two-tailed *t* test with significance reported as a *P* value of ≤ 0.05 .

RESULTS

Data points for the six different silicone phantoms generated from compression of the samples were fitted to the stress/compression curves as illustrated in Fig. 6. Increasing stress is applied to the samples (y-axis) in to produce a given compression ratio. Series 1 is thus the hardest material (least compliance) and series 6 is the softest silicone phantom (greatest compliance). This serves as the basis for comparing the subjective rating of the silicone phantoms by the 10 subjects to the correct order of varying compliance.

Fig. 7 depicts the mean of the squared errors of the subjective order of compliance (compared to the known order) for surgeons and nonsurgeons for each of the three tool types. Although the nonsurgeons (electrical engineers with extensive experience in haptic technology) appeared to have fewer errors in determining the correct order (lower error score) than the surgeons, this difference was not significant ($P > 0.05$).

Fig. 7 shows pooled data from both groups ($n = 10$ subjects, surgeons and nonsurgeons) on the subjective rating of sample compliance. The force-feedback Babcock grasper yielded improved force feedback when compared with the standard nondisposable Babcock grasper ($P \leq 0.05$) with a tool tip of identical mechanics and surface area. The human hand was significantly better ($P \leq 0.05$) than the other two in determining the correct order of sample compliance.

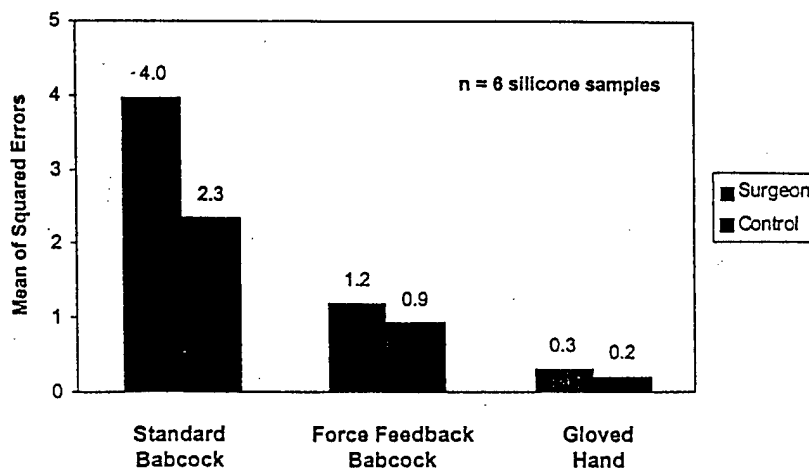


Fig. 7. Palpation experiments of six silicone phantoms by surgeons and nonsurgeons (*control*). Three different tools (standard laparoscopic Babcock grasper, force-feedback Babcock grasper, and gloved hand) were used. No significant difference is observed between the two (surgeon and nonsurgeon) groups.

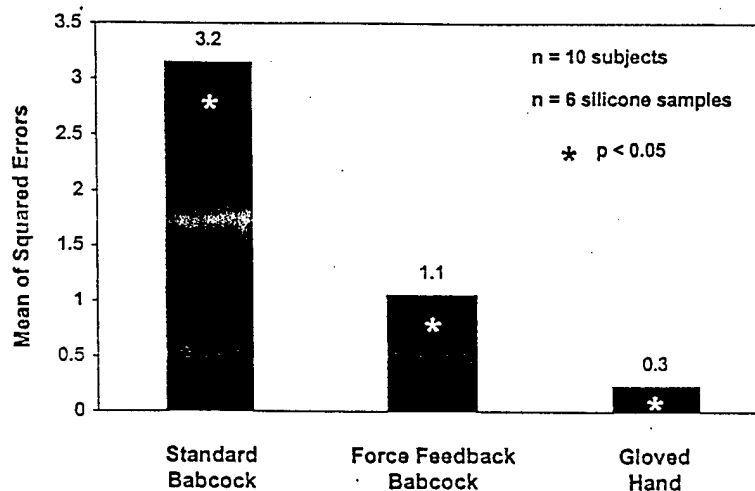


Fig. 8. Silicone phantom palpation error scores. Pooled data (surgeon and nonsurgeon) from palpation experiments on the six silicone phantoms. Significant differences are observed based on the palpation tool used.

DISCUSSION

Endoscopic-based operations continue to expand into new areas of surgery. Operations that were once thought possible to complete only by means of an open approach are now being performed commonly via an endoscopic approach.¹³ In order for an endoscopic approach to surgery to be beneficial, it must also be safe. Although most endoscopic surgery proceeds without incident, there are reports of injury to organs resulting in significant complications.¹³ Surgeons need to rely on visual cues to compensate for lack of depth perception and poor haptic feedback secondary to reliance on long instruments to perform the operation.

To potentially improve the force feedback during endoscopic surgery, we have designed and tested a prototype surgical endoscopic instrument that has the advantage of easy tool tip interchangeability with existing marketed tool tips. Although the prototype design is bulky, the intracavitary portion of the tool is identical to that of conventional endoscopic instruments.

As depicted in Fig. 8, the human hand is superior to both the force-feedback Babcock instrument and the standard Babcock endoscopic grasper at determining the correct order of silicone phantom compliance. This is not surprising given the fact that the human hand is a highly complex diagnostic tool and has

multiple haptic properties (sense of position, proprioception, temperature, texture sensation). What is surprising is the degree of haptic sensation improvement of the force-feedback Babcock vs. the standard Babcock grasper. Although the time to complete each rating of the six samples was not measured, there was a noticeable time difference to complete each subjective rating by all subjects (human hand, force feedback grasper, standard grasper with increasing time requirements, respectively). This further underscores the differences in difficulty in determining the correct order of phantom compliance for each tool type.

The premise that improved force feedback will result in less tissue injury during endoscopic operations is difficult to assess since improved force feedback is not yet available to the endoscopic surgeon. A similar premise is that three-dimensional imagery may produce increased efficiency in endoscopic operations.^{15,16} It seems apparent that visual cues are very powerful in filling in the gaps of visual and haptic deficits in endoscopic surgery, but these "compensations" may not be necessarily accurate or safe.¹⁴⁻¹⁶ Until improved force-feedback capabilities can be made available for testing, we will not know if improved haptic feedback will result in fewer injuries to soft tissues. In addition, it is not known how much force and torque applied to a given tissue by laparoscopic operations will result in tissue injury, either reversible or irreversible injury. This aspect is currently undergoing study in our laboratory.

The inability to palpate tissues accurately during endoscopic surgery because of inadequate force feedback undoubtedly results in loss of diagnostic information. As surgeons of the open surgery era, we have been spoiled by the luxury of the human hand to supply this diagnostic information. Common examples would include hand palpation of common bile duct stones, lung and liver nodules, and intestinal masses. As endoscopic surgery further displaces open surgery as the standard of care, we will be additionally handicapped in our diagnostic intraoperative capabilities. Although endoscopic ultrasound imaging has excellent potential for bridging some of this gap of intraoperative diagnostic limitations,^{19,20} it does not work well on hollow organs because of artifacts of shadowing from air/tissue interfaces. Future generations of force-feedback devices will surely improve in their compactness and fidelity of information. Furthermore, we have previously reported with a computerized grasper that normal biologic organs have characteristic force profiles¹⁷ based on their intrinsic tissue properties. If the force profile is not "normal" for a given biologic tissue, then this could represent tissue inflammation, fibrosis, foreign body, or cancer, thus improving diagnostic yield.

Another potential application of this technology is in the area of tissue protection. If the degree of forces and torques that result in tissue injury can be determined (work in progress), then "smart endoscopic force-feedback instruments" can be developed that will apply force/torque limitations resulting in tissue protection from iatrogenic operative tissue injury.

CONCLUSION

Although minimally invasive surgery techniques have brought forth astounding changes in surgical care, with benefit to all participants who provide and receive health care, we need to continue to strive to make operations safer, more efficient, and with fewer complications. We have demonstrated that haptic feedback can be potentially improved during minimally invasive surgery. Whether this will translate into fewer episodes of tissue injury and improved diagnostic capabilities remains unclear. However, the first step is to be able to make this technology available for testing. The ultimate goal is to perform operations which are performed more efficiently, less invasively, and with fewer complications to patients.

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Discussion

Dr. L. W. Traverso (Seattle, Wash.). Of the three common laparoscopic procedures we do—cholecystectomy, Nissen fundoplication, or appendectomy—which do you think will be helped the most by this technology?

Dr. M. MacFarlane. We certainly perform many laparoscopic Nissen funduplications and I can say that we see a fair number of serosal tears, although perforations are very rare in our hands. So I think the more advanced cases are going to be the best application for this technology.

Dr. B. Schirmer (Charlottesville, Va.). This is a very interesting concept, but I am a little concerned about whether it is going to be generally applicable. Do you see this as being something that every grasper will be equipped with eventually and at what cost? Will equipping these graspers be cost-effective? Second, if you were to design just one type of instrument, in which specific situations would you use it? Ultrasound imaging does very well for solid organs—would you, for example, want to use it in the colon to palpate lesions? What else do you envision for this application?

Dr. MacFarlane. Laparoscopic ultrasound is a wonderful modality. It is very sensitive, especially on solid organs, so I think the grasper would be most applicable, in terms of diagnostic information, on hollow organs. Second, if the

cost of new technology were the sole determining factor, we would not have any new technology. Until this technology is developed further, I do not think we will know the answers to your questions. In its current prototype form, our instrument is very bulky. It needs to have stronger actuator forces to carry out the steps in the operation that are performed.

Dr. N. Soper (St. Louis, Mo.). You are at the low end of the development of this instrument and I think there is a long way to go. In terms of the bulkiness of the instrument as currently designed, it looks like there will need to be something at the actuator end, which will bulk up the end that is going through these little incisions we are going to make. Is there any possibility of placing that actuator back out, near the handpiece for instance, rather than down at the tip?

Dr. MacFarlane. The actuator itself does not limit our capabilities inside the abdomen because the length of the intra-abdominal portion is identical to what is currently used in laparoscopic surgery. As with all prototypes, future models will become more compact and more efficient. I cannot currently estimate what the final product will look like.

Force Controlled and Teleoperated Endoscopic Grasper for Minimally Invasive Surgery - Experimental Performance Evaluation

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Abstract

Minimally Invasive Surgery (MIS) generates new user interfaces which create visual and haptic distortion when compared to traditional surgery. In order to regain the tactile and kinesthetic information that is lost, a computerized Force Feedback Endoscopic surgical Grasper (FREG) was developed with computer control and a haptic user interface. The system uses standard unmodified grasper shafts and tips. The FREG can control grasping forces either by surgeon teleoperation control, or under software control. The FREG performance was evaluated using an automated palpation function (programmed series of compressions) in which the grasper measures mechanical properties of the grasped materials. The material parameters obtained from measurements showed the ability of the FREG to discriminate between different types of normal soft tissues (small bowel, lung, spleen, liver, colon and stomach) and different kinds of artificial soft tissue replication materials (latex/Silicone) for simulation purposes. In addition, subjective tests of ranking stiffness of silicone materials using the FREG teleoperation mode showed significant improvement in the performance compared to the standard endoscopic grasper. Moreover, the FREG performance was closer to the performance of the human hand than the standard endoscopic grasper. The FREG as a tool incorporating the force feedback teleoperation technology may provide the basis for application in telesurgery, clinical endoscopic surgery, surgical training, and research.

1. Introduction

Minimally Invasive Surgery (MIS) is a new technique in which a surgeon operates with specially designed surgical tools through access ports requiring incisions of about 1cm in size. This limits the surgical trauma to tissues, decreases the pain that the patients experience and results in a significant shortened recovery period. MIS technology generates two new user interfaces (i) the monitor which gives the surgeon a 2D visual feedback of the internal anatomy, and (ii) the MIS surgical instruments which are inserted through the access ports. The instruments enable the surgeon to manipulate the internal organs by using finger loops outside the patient's body linked to the tool tip via a long tube/shaft including internal mechanism. Despite the benefits of MIS, this technique has some disadvantages due to the two new interfaces. The monitor, as a visual interface, reduces the surgeon's perception from a 3D to a 2D view of the anatomy. Furthermore, the MIS instruments limit the surgeon's ability to gain the diagnostic information about the tissue being manipulated, as opposed to traditional surgery in which the surgeon examines the tissue by touching it directly with the hands. Moreover, due to the internal friction and backlash of the mechanism of MIS instruments the ability to perceive information by palpating tissues and organs is significantly reduced. Two components of the palpation information are tactile and kinesthetic information. Their combined use is referred to as haptic perception.

Previous work has explored manual driven material palpation with regular instrumented endoscopic grasper [1], and automated palpation with an external robot [2]. Tactile sensors have been applied to endoscopic graspers which are coupled to tactile displays [3,4]. These systems aim to enable the surgeon to discriminate textural or time varying features of the tissue via endoscopic tools. An instrumented Babcock grasper, which measures forces and torques at the tool-tissue interface has been reported, but does not measure or control grasping force [5]. In addition, teleoperation robots [6,7,8] and simulators implementing virtual reality with a force feedback haptic device have been developed [8, 9].

The importance of haptic feedback to safely perform surgery is unclear. Psychophysical experiments investigating the ability of human to tactually discriminate the softness of objects showed that whereas tactile information was sufficient for rubber specimens both tactile and kinesthetic information was found necessary for spring cells [10]. Although color, texture, and visible aspects of tissue deformation in the surgical field convey important anatomic information, palpation may be critical in identifying otherwise obscure tissue planes, arterial pulsations, and regions of tissue thickening that may signify pathology such as infection or cancer. Safe tissue handling requires tissue manipulation that is both secure and non-damaging. Much of the art of surgery and the implicit learning curve for traditional surgical technique depend on training to refine and educate the sense of touch. Training for endoscopic surgery is even more difficult because of the remote nature of Videoendoscopic tissue manipulation. Recent literature emphasizes the importance of tactile feedback for accurate targeting of primary [11, 12, 13] and

metastatic cancer [14, 15, 16, 17] and identifying therapeutic margins for curative resection [12, 18, 19].

The loss of palpation for localization may seriously limit the efficacy and safety of minimally invasive treatment in some operative fields [17]. The present study aims to develop and characterize a grasper capable of restoring a degree of kinesthetic information to the surgeon about the tissue being grasped. The following goals were laid out: (i) improve the ability of the endoscopic surgeon to feel mechanical properties of tissues such as compliance, (ii) make minimal changes to the form and function of existing surgical graspers to reduce cost, complexity, and certification difficulties (i.e. avoid adding sensors and wiring to the tool tip), (iii) take advantage of the declining cost of computer control.

This study is focused on two aspects of using the FREG: (i) objective aspect in which the FREG was used in an automatic mode for testing mechanical characteristics of soft tissue and viscoelastic material replications and, (ii) subjective aspect (psychophysics) in which the FREG was operated in bilateral force feedback mode examining the performance of test operators ranking materials according to their stiffness and comparing to the performance achieved by using a regular grasper and a gloved hand.

2. Methods

2.1 Force Feedback Endoscopic Grasper (FREG)

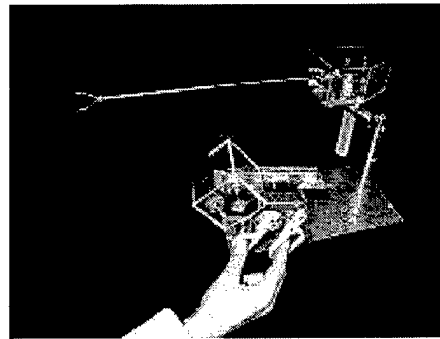
2.1.1 System Overview

The Force Feedback endoscopic Grasper - FREG (Fig. 1) incorporates teleoperation technology into an existing, reusable, endoscopic grasper (Fig. 2) for minimally invasive surgery [20]. The FREG system includes two subsystems. The master and the slave each consist of an actuator and a position encoder. The tool tip, pull/push rod and tube is mounted on the slave subsystem which is inserted into the patient's body through an access port. The proximal end of the instrument tube is clamped to a supporting post of the slave. The pull/push rod operating the tool tip (jaws) is linked to the electromagnetic actuator via a ball and socket joint. The two finger loops (user interface) of the grasper are mounted on the master subsystem. The distal finger loop is connected to an actuator/encoder pair identical to those on the tool shaft enabling the surgeon to control the tool tip.

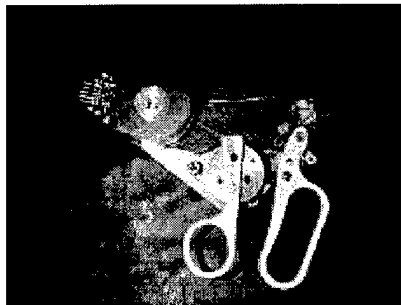
To increase sensing resolution, the encoder wheels are connected to the actuation axes via pulleys and a kevlar drive belt having a multiplication ratio of 1:3.6. As a consequence, both master and slave position sensors have 1400 quadrature position counts over the full 0.6 radian (34.4°) motion range. The FREG actuators are flat coil actuators modified from hard disk drive head positioning actuators. Hard disk drive head actuators have many advantages for precision robotics and force feedback devices [21]. Actuators taken

directly from 5.25 inch (133mm) hard drives with a maximum torque of 0.1NM at 2.0 A (based on steady state coil temperature of 93° C) did not produce convincing subjective grasping sensations. The actuator magnets were replaced with custom made Nd-Fe-B magnets having approximately triple the energy product of the Al-Ni-Co magnets used in the disk drive actuator [22]. The coil and bearing assembly was retained. To realize the full flux increase from the new magnets, new frames were built from high permeability iron to prevent backing iron saturation. The new actuator magnets and frames increased the torque output to 0.3 NM, but preserved the desirable qualities of low torque ripple, low friction, and low back-driving inertia.

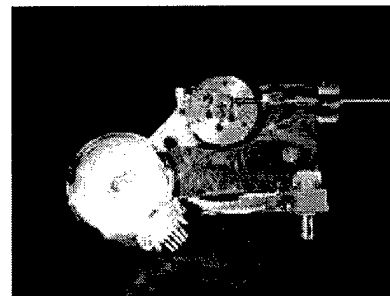
(a)



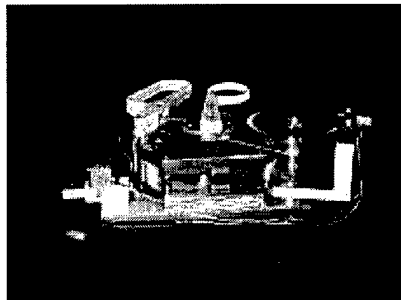
(b)



(d)



(c)



(e)

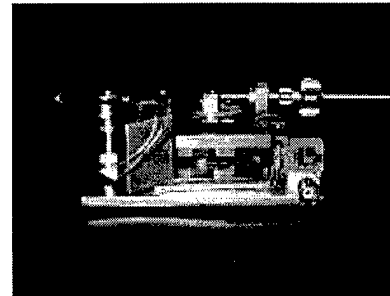


Figure 1: The Force Feedback Endoscopic Grasper - FREG - (a) System overview; (b) Master - Front view (c) Master - Top view; (d) Slave - Front view; (e) Slave - Top view

The Laparoscopic instrument used in these experiments is a stainless steel Babcock grasper (Carl Storz Inc., model # 30420 BL) with a square jaw grasping surface area measuring 9.4 x 8.5 mm (Fig. 2). The tool shaft is 5 mm in diameter and 38 cm long from the proximal attachment to the instrument tip. The shaft and mount allow 360 degree rotation of the tool about its long axis. This system allows easy change of shaft length, diameter, and tool tip conformations. Laparoscopic tools compatible with the mounting system are readily available from various manufacturers.

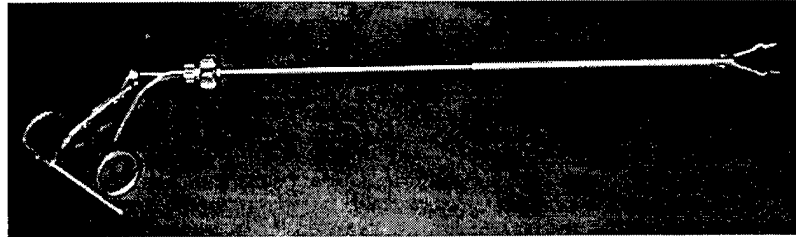


Figure 2: Endoscopic grasper with a Babcock tip- Carl Storz model No. 30420 BL

2.1.2 Control

The control system supports two modes of operation (i) bi-lateral force feedback - teleoperation, and (ii) programmed automatic grasping (palpation) operation for tissue characterization (Fig. 3). Proportional-derivative (PD) controllers were designed for both the master and slave using a linear dynamic model of the device and conventional control techniques [24]. Integral feedback is not desirable in position error based force feedback control because it creates a time varying force feedback under conditions of steady state contact. The force feedback controller is based on the well known bi-lateral, position error based, teleoperation system [25]. In this design, the measured position of each side serves as the reference position input for the other.

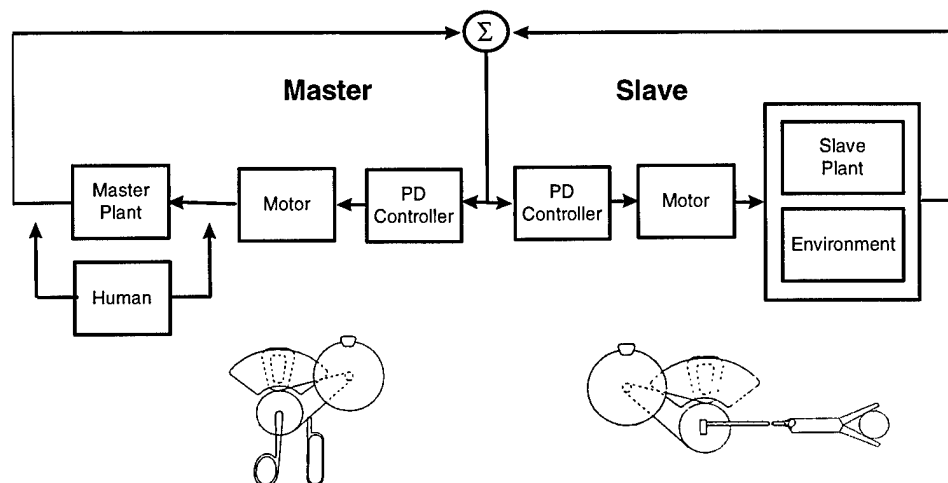


Figure 3: FREG Bi-lateral Force feedback (Teleoperation) control scheme

A desirable quality of force feedback system is a high effective stiffness between master and slave sides. In the position error based architecture, this requirement can be translated into the need for a high value of the proportional feedback gain, K_p [25]. An additional controller design constraint is introduced from the actuator limit of 0.3Nm maximum torque. Experience shows that users feel a subjective loss of contact sensations when a force feedback device saturates at its maximum force output. There is thus a trade-off between K_p and the deflection at which saturation occurs. For high values of K_p , the user will feel high effective stiffness, but saturation will occur at relatively smaller position errors. The K_p value was selected such that the position error corresponding to the actuator saturation point is set at one quarter of the motion travel range (Eq. 1).

$$K_p = \frac{I_{\max}}{0.16} = 12.6 \quad (1)$$

The remaining parameter K_d was determined by placing the dominant closed loop pole for an 8 ms settling time constant and a damping ratio of 0.5. For the slave, this design method resulted in an unstable controller, possibly because of backlash in its mechanism. An acceptable controller was recomputed with a lower initial K_p value. The resulting gains are given in Table 1.

	K_p [Nm/rad]	K_d [Nm sec/rad]
Master	12.6	0.05
Slave	9.6	0.04

Table 1: Controller Parameters

2.1.3 Mechanism Analysis

The endoscopic grasper has a unique mechanism to transfer the position and moments applied by the surgeon on the finger loops to the tool at the tip which is grasping the tissue. The following static analysis takes into account only the grasper geometry and not the system dynamics and its friction. Fig. 4 shows a scheme of the endoscopic grasper internal mechanism in two typical positions: (i) tool tip jaws closed (reference position) (ii) tool tip jaws in an intermediate position. Given the geometry defined in Fig. 4, the finger loop angle (θ) as a function of the tool tip jaws angle (ϕ) is defined by Eq. 2. The transfer function between the moment applied on the finger loops relative to their joint (T_H), and the moment developed at the tool tip jaws relative to their joint (T_T) is defined by Eq. 3. Using the numerical geometry dimensions of the Babcock grasper (Carl Storz Inc., model # 30420 BL) the transfer functions (Eq. 2, 3) are plotted in Fig. 5a b.

$$\theta = \tan^{-1} \left\{ \frac{1}{R} \left[L_0 - L_1 \sqrt{1 - \left(\frac{L_2}{L_1} \sin \left(\beta_0 + \frac{\varphi}{2} \right) \right)} - L_2 \cos \left(\beta_0 + \frac{\varphi}{2} \right) \right] \right\} \quad (2)$$

$$\frac{T_L}{T_H} = \frac{L_2 \cos \theta}{2R} \left[\sin \left(\beta_0 + \frac{\varphi}{2} \right) + \frac{\frac{L_2}{2L_1} \sin \left(\beta_0 + \frac{\varphi}{2} \right)}{\sqrt{1 - \frac{L_2}{L_1} \sin \left(\beta_0 + \frac{\varphi}{2} \right)}} \right] \quad (3)$$

An ideal transfer function of the endoscopic grasper would be a linear transfer with a gain of one for the hand to tip tool position (Fig. 5a), and a constant value of the tool tip moment to handle moment ratio as a function of the handle position (Fig 5b). However using the geometry dimension of the grasper under study shows a gain of 1.1 between the handle position and the tool tip position (Fig 5a) and nonlinear moment ratio between the handle and the tool tip as a function of the handle angle (Fig 5b). The grasper mechanism moment/position transfer function might be another reason for the kinesthetic distortion that exists in endoscopic tools.

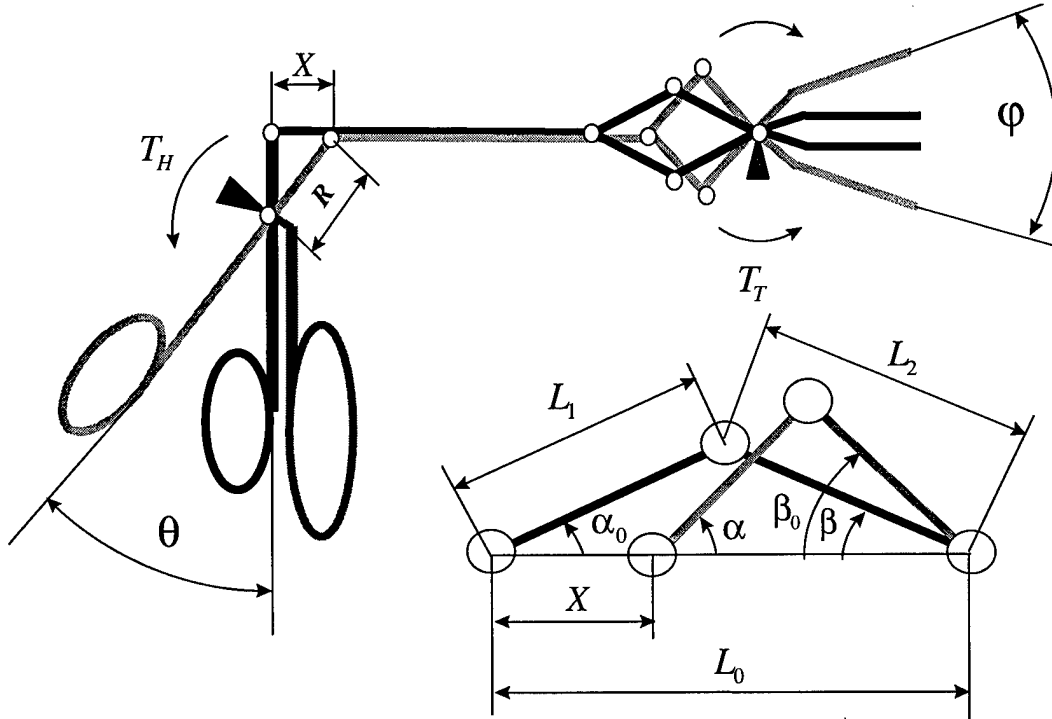


Figure 4: Endoscopic grasper - Schematic diagram of the mechanism in two typical positions

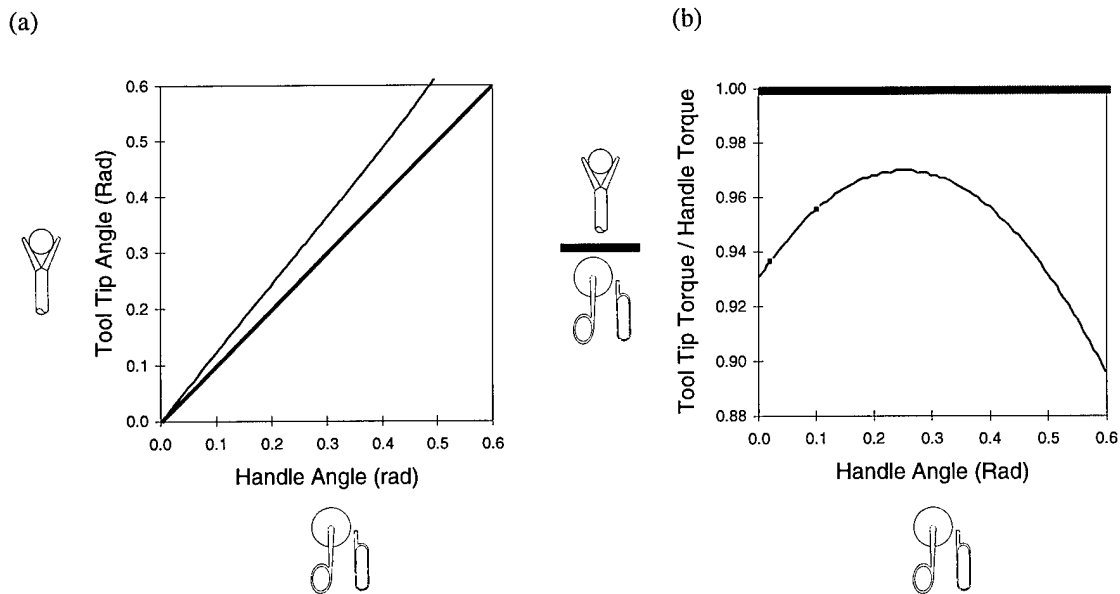


Figure 5: Endoscopic grasper transfer functions (Thick line - Linear transfer function; Thin line - Actual transfer function): (a) Tool tip angle as a function of handle angle; (b) Tool tip moment - handle moment ratio as a function of handle angle.

2.2 Materials

Two type of materials were used in order to evaluate the FREG performance: (i) soft tissue (Pig internal organs - Small Bowel, Spleen, Colon, Stomach, Liver, and Lung), (ii) latex and Silicone materials. The soft tissue, latex and Silicone material were used for evaluation of their mechanical characteristics using the FREG in automatic palpation mode (objective protocol - section 2.3.1). In addition the Silicone materials were also used to test the FREG performance in teleoperation mode compare to other tools in ranking the materials stiffness by test operators (subjective protocol - section 2.3.2). Table 2 summaries the materials, tool usage and mode of operation in each type of experimental protocol.

Protocol	Type of Material	No. of Specimens	Tool	Operation Mode
Objective	Soft Tissue	6	FREG	Automatic
	Latex	5	FREG	Automatic
	Silicone	8	FREG	Automatic
			Indenter	Automatic
Subjective	Silicone	6	FREG	Teleoperation
			Grasper	Manuel
			Hand	Manuel

Table 2: Summary of tools and materials used in the experimental protocol

The latex and the Silicone materials were manufactured by Simulab Corporation. For the purpose of further discussion, latex materials were designated L1, L2, L3, L4, L5. All the latex materials were shaped in the same cylindrical form with a diameter of 13 mm and a length of 45 mm. The above designation referred to each material based on its stiffness, where L1 is the softest material and L5 is the stiffest material. L1 to L4 can be considered viscoelastic materials representing artificial replication of soft tissues, while L5 can be defined as a solid which exhibits the upper limit of physiological stiffness, and can simulate the bone.

The Silicone materials compliance characteristics were controlled by the percentage (weight) of catalyst used during manufacturing. A set of 8 materials were thus obtained (0%, 5%, 10%, 15%, 20%, 25%, 30%, 35% of catalyst). All the Silicone materials were shaped as a cylinder with a diameter of 14.7 mm and a length of 150 mm with the same color and texture.

2.3 Experimental Protocol and Data Analysis

Two types of protocols were defined and tested in order to evaluate the performance of the FREG system in its two modes of operation. The objective experiment focused on the biomechanical characteristics of soft tissue and viscoelastic material replication using the FREG in an automatic mode. In the subjective experiment, test operators used the FREG in a bilateral mode using a protocol which examined the ability to rank material stiffness, relative to the performance achieved by a standard endoscopic grasper tool and the human hand (Table 2).

2.3.1 Objective Experiment

The FREG is controlled by software running in real-time mode on a PC. This feature allows automated grasping and palpation functions to be implemented in software. Using this function, the deflections and forces being applied on the tissues are measured in order to extract information about tissue mechanical properties. The aim of the protocol was to measure mechanical characteristics of biological and artificial soft tissue materials with the endoscopic grasper tool.

A material can be defined by its constitutive equation, however those equations can only be determined by experimental methods [26]. Previous studies analyzing soft-tissue were focused mainly on testing tissues under uniaxial tensile conditions using the Quasi-Linear Viscoelasticity theory - QLV [26]. Selected tissues, which are inherently under physiological compression conditions (e.g. bone, cartilage, intervertebral disc), have been studied under uniaxial compression using the biphasic viscoelastic theory [27].

A full characterization of soft-tissue and latex as viscoelastic materials requires an extensive experimental data-base which includes the material time domain response (Creep and Relaxation) and its frequency response (Complex Modulus). Fung [26] in his QLV

theory suggested that if a step increase in elongation is imposed on the specimen, the stress developed will be a function of time (t) as well as of the material stretch ratio (λ). The history of the stress response, called the relaxation function ($K(\lambda, t)$), is assumed to be of the form

$$K(\lambda, t) = G(t) * T(\lambda) \quad (4)$$

in which $G(t)$ is a normalized function of time, called the reduced relaxation function and $T(\lambda)$ is a function of the stretch ratio alone, called the elastic response. Fung [26] proposed Eq. 5a for defining the elastic response of the material under tension conditions.

In our study the integration constant (C) in Eq. 5a was determined by using the initial condition in a natural state in which, by definition, $T = 0$ when $\lambda = 1$. Moreover, Eq. 5a was modified to define the elastic response in compression condition (Eq. 5b) by multiplying the right hand side of Eq. 5a by -1. We define the stress in compression condition with a negative sign compare to tension, and we substitute the definition of tensile strain $\epsilon_T = (L - L_0)/L_0 = \lambda - 1$ with the definition of compressive strain $\epsilon_C = (L_0 - L)/L_0 = 1 - \lambda$.

$$T_T(\lambda) = Ce^{\alpha\lambda} - \beta \quad (a) \quad (5)$$

$$T_C(\lambda) = \beta(1 - e^{\alpha(1-\lambda)}) \quad (b)$$

$$T = \frac{F}{A_0} \quad (6)$$

$$\lambda = \frac{L}{L_0} \quad (7)$$

$$K(\lambda) = \frac{dT}{d\lambda} = \alpha\beta e^{\alpha(1-\lambda)} \quad (8)$$

where α and β are parameters, λ is the compression length-ratio, T_T is the uniaxial tension stress, T_C is the uniaxial compression stress, F is the force applied on the specimen by the grasper, A_0 is the grasper contact cross section area, L is the length of the material compressed by the load, L_0 is the length of the material at zero load, and K is the material stiffness.

To perform the automated palpation function, the slave position controller was driven by a sinusoidal displacement command while recording position, position error, and its motor torque. Three cycles of a 1 Hz sinusoidal displacement were applied as the command position input to the slave controller. The amplitude of the sinusoid corresponded to full opening and closing of the jaws (0.6 rad). During the automated palpation the slave received its control commands directly from the computer and the master was

disconnected. The PC recorded the slave torque and angular displacement for three seconds at a sampling frequency of 1 KHz.

Torque vs. displacement data were first isolated in time to the segment involving initial contact and compressive displacement. Compared to other types of surgical grasping instruments, the geometry of the Babcock tool suggests that it creates a relatively uniform stress distribution under the contact sites. The torque- displacement data measured at the handle were transformed to the uniaxial compression stress-length ratio using the geometry of the slave. Then, the stress length-ratio data were fitted, using the Least-Square method, with Eq. 5b.

The device performance was evaluated with in-vivo palpation of pig internal organs (Small Bowel, Spleen, Colon, Stomach, Liver, and Lung). Tissues were inserted in the jaws with the tool hand held. Three experimental sessions were performed compressing the tissue in three different locations. Each experimental session included three sinusoidal compression cycles in the same locations. Thus, a total of 9 Stress length-ratio data sets were examined for each material. Protocols for anesthetic management, euthanasia, and survival procedures were reviewed and approved by the Animal Care Committee of the University of Washington and the Animal Use Review Division of the U.S. Army Veterinary Corps. In addition to the soft-tissue the same experimental protocol was used to test the latex and silicone materials.

In order to independently test the materials with a method which is less dependent on the testing tool type, the stress compression length-ratio of the Silicone materials were measured by a parallel surface compression test. In this testing method the Silicone materials were placed on a flat surface plate and a metal cylinder indenter with a diameter of 7 mm and a flat contact area which was parallel to plate was penetrated to the materials. The indenter was moved against the plate along the Silicone material diameter applying compression conditions while measuring the force and deflection along the line of action. No additional boundary conditions except the flat surface and the indenter restricted the materials from expending freely in all other direction.

2.3.2 Subjective Experiment

An objective method to evaluate the FREG performance in discriminating between the stress length-ratio characteristics of soft tissue and viscoelastic materials was described in the previous section. However, the endoscopic graspers are usually manipulated by a human. This situation raises the question regarding the psychophysical aspects of the FREG performance. The experimental protocol was designed to examine the operator ability to rank a group of materials according to their stiffness using three tools: (i) the FREG in a force feedback bi-lateral mode (ii) a regular endoscopic grasper, commonly used in MIS, and (iii) touching the materials with the hand (with a latex glove) as traditionally done in open surgery.

Ten test operators divided into two groups (surgeon group and control group) were asked to rank eight materials according to their stiffness using three tools. In order to avoid the fatigue and learning effects, each test operator performed the experiment using the three tools in different order. A typical experiment, using one of the tools mentioned above, began with a 2-5 min learning period in which the materials were presented to the test operators in the correct order of increasing or decreasing stiffness. The test operators were allowed to learn by using the current tool, the differences in stiffness between the Silicone samples. During the testing session, the materials were presented to the test operators in random order (keeping the same random order for all the test operators). The test operators were then instructed to rank the materials in regard to stiffness four times (two times in increasing stiffness order; and two times in decreasing stiffness order). The experiment was performed in such a way that the test operators could not see either their hand nor the tool tip, in order to avoid visual feedback of material deformation. Using this method, the test operator's material ranking was based solely on their haptics sense (touch). The test operators wore a single pair of surgical latex gloves during the experiment to simulate operating room conditions. In the case of ranking the material stiffness with the hand, the test operators were instructed to apply only compression forces along the material's diameter (cylinder shape) rather than shear forces. The ranking methodology that the test operators were instructed to follow was to build a subgroup, by excluding out of the whole group 2-4 materials with the highest or lowest stiffness and then to rank the materials in this group. If the test operator was unsure regarding the stiffness of the last material in the subgroup, it was possible to return it to the initial group.

The statistical null hypothesis (H_0) was that all the tools (standard endoscopic grasper, FREG, and hand) are equal in their performance for ranking material stiffness. The index of performances was calculated as the Mean Squared Error (MSE) of the estimated ranking (ER) relative to the correct ranking (CR) - Eq. 9.

$$MSE = \frac{\sum_{i=1}^n (ER - CR)^2}{n} \quad (9)$$

For the six materials ($n=6$ - Eq. 9) used in the subjective experiment, the MSE value has a range of $< 0, 11.66 >$. When a test operator ranks the materials exactly according to their correct stiffness order the MSE has a value of 0. The upper boundary of MSE value of 11.6 can be achieved when the test operator ranks the material in the opposite manner. Two way analysis of variance (2D ANOVA) was used to analyze the differences between instruments (Regular endoscopic grasper, FREG, and hand) and between groups (surgeon group and control group).

3. Results

3.1 Objective Experiment

The stress length-ratio curves for the compression phase of each material grasping (Pig internal organ soft tissues, latex, and Silicone materials) were fitted with the exponential function (Eq. 5b) using the least squares method. Typical measured data and the corresponding curve fit are plotted in Fig 6 for each material. Note, that for presentation purposes, minus signs indicating compression stresses were omitted. Since the software generated three close/open cycles, there were three compression tests recorded 1 second apart for each grasp. Tissues typically got stiffer in the second and third compression of each sequence.

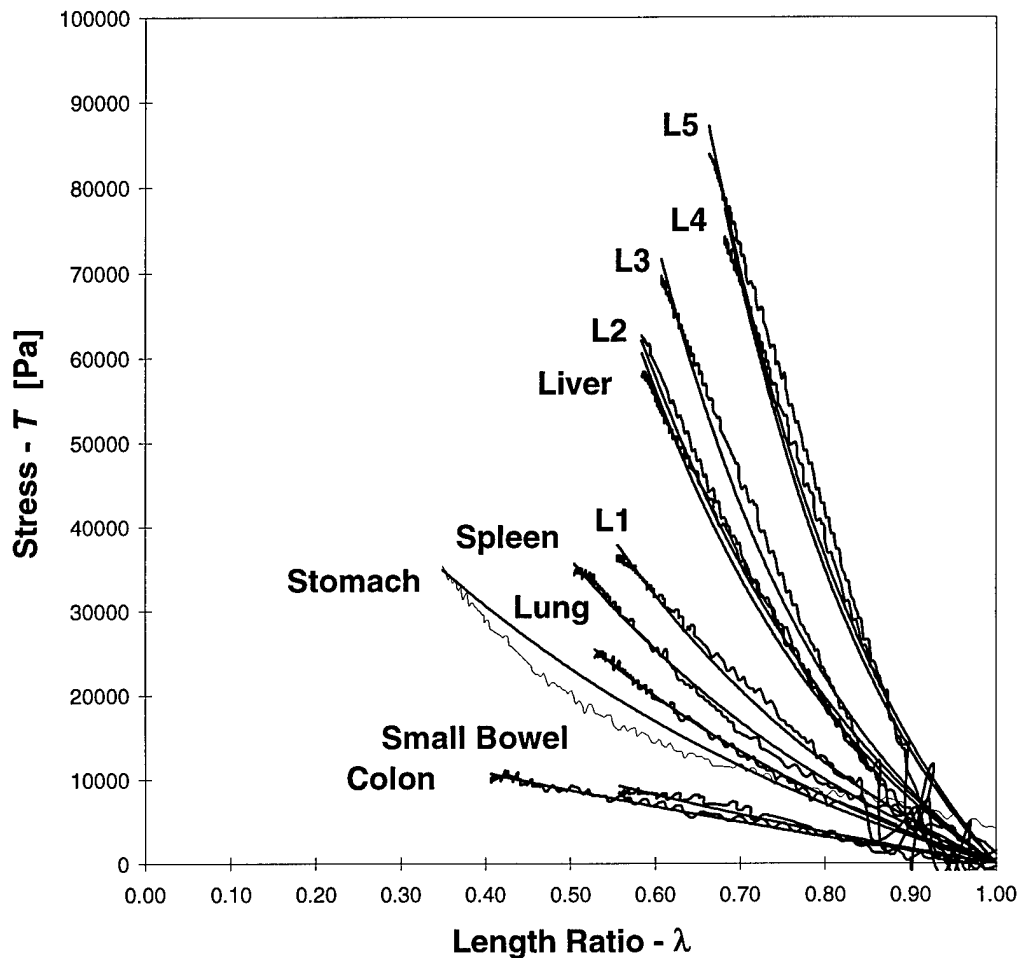


Figure 6: Typical uniaxial compression stress as a function of length ratio and the corresponding exponential curve fit for different Pig organs and latex materials.

The quality of the numerical fit was verified by using the correlation ratio factor R^2 . The computed R^2 values were typically very close to one ($R^2 > 0.999$), indicating very high quality of fit between Equation 5b and the experimental data. Two exceptions with relatively lower R^2 values ($R^2 > 0.99$) were colon and stomach tissue. These tissues exhibited a different type of compression characteristics especially at lower compression length ratios.

The resulting parameters of the exponential fit α and β as computed from the FREG's measurements were displayed as a scatter plot for the pig soft tissues and for the latex and Silicone materials (Fig. 7). Generally speaking, stiff materials have high values of α and β , and vice versa. In some cases, as the stiffness of the material increases, β increases whereas α may decrease (as happened for the Silicone materials tested using the indenter technique - Fig. 8). However, the product $\alpha \cdot \beta$ always increases as the material stiffness increases, indicated by Eq. 8. Data points of α and β formed into clusters for each material that was tested. Each cluster consists of nine data points. Rectangles, defined by the standard deviations computed from the organ data clusters, did not overlap except for lung and spleen. These variances are partly due to the variation in stiffening of tissues under repeated compression as described above.

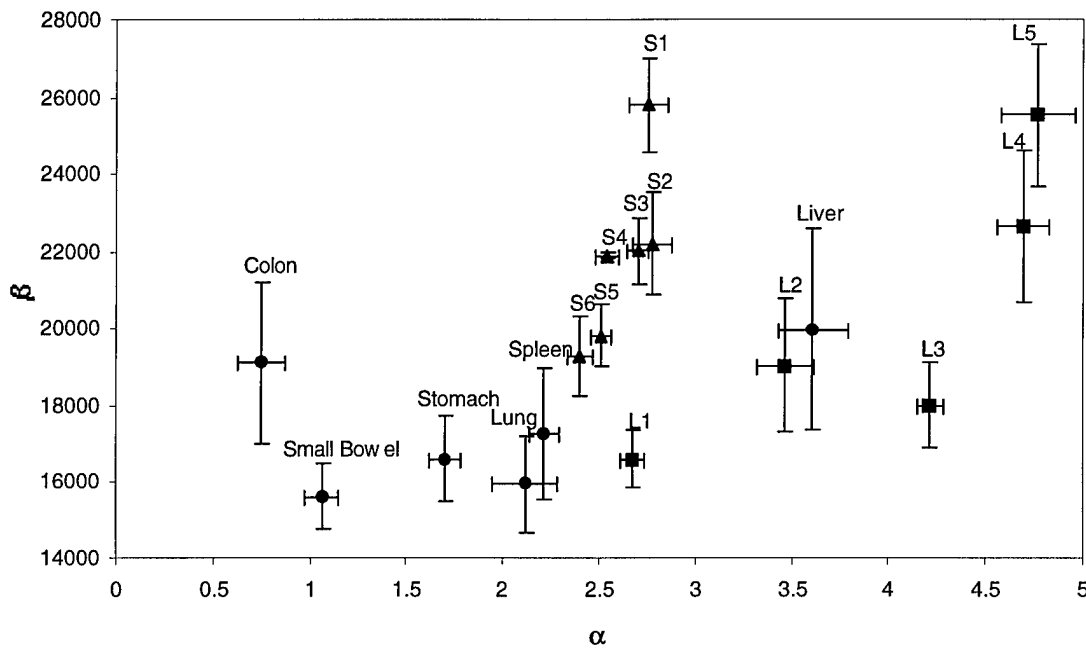


Figure 7: Scatter plot of the biomechanical parameters α and β for different Pig soft tissues, latex (L1-L5) and Silicone (S1-S6) materials.

In addition to the measurements performed by the FREG, the 8 Silicone materials were tested using the indenter experimental protocol. Stress-Strain plots of the 8 silicone materials indicated that from the mechanical point of view 3 of the 8 materials (20%, 25% and 30%) had the same stress-strain curves, so that it was impossible to distinguish

between their stiffness with the tools used in the subjective experiment. For the purpose of analyzing the data, the three materials with the similar mechanical characteristics were lumped into one. The exponential fit parameters α and β obtained by indenter testing method have different values ($\alpha < 4.76, 5.57, 5.89, 5.85, 6.14, 7.07 >$ $\beta < 6300, 3034, 1830, 1379, 941, 380 >$ for (S1- S6) compared to those measured by the FREG (Fig. 7, Fig. 8). This phenomena can be explained by the FREG inherent stiffness due to its structure and internal mechanism. The FREG stiffness most probably was measured as part of the materials stiffness. However, the relationship between the material parameters multiplication $\alpha \cdot \beta$ and the material stiffness remained the same for the two testing methods. Moreover, since the 6 Silicone materials are evenly graded as measured by stress length-ratio characteristics (Fig. 8), this makes them ideal for the purpose of the subjective testing experiment.

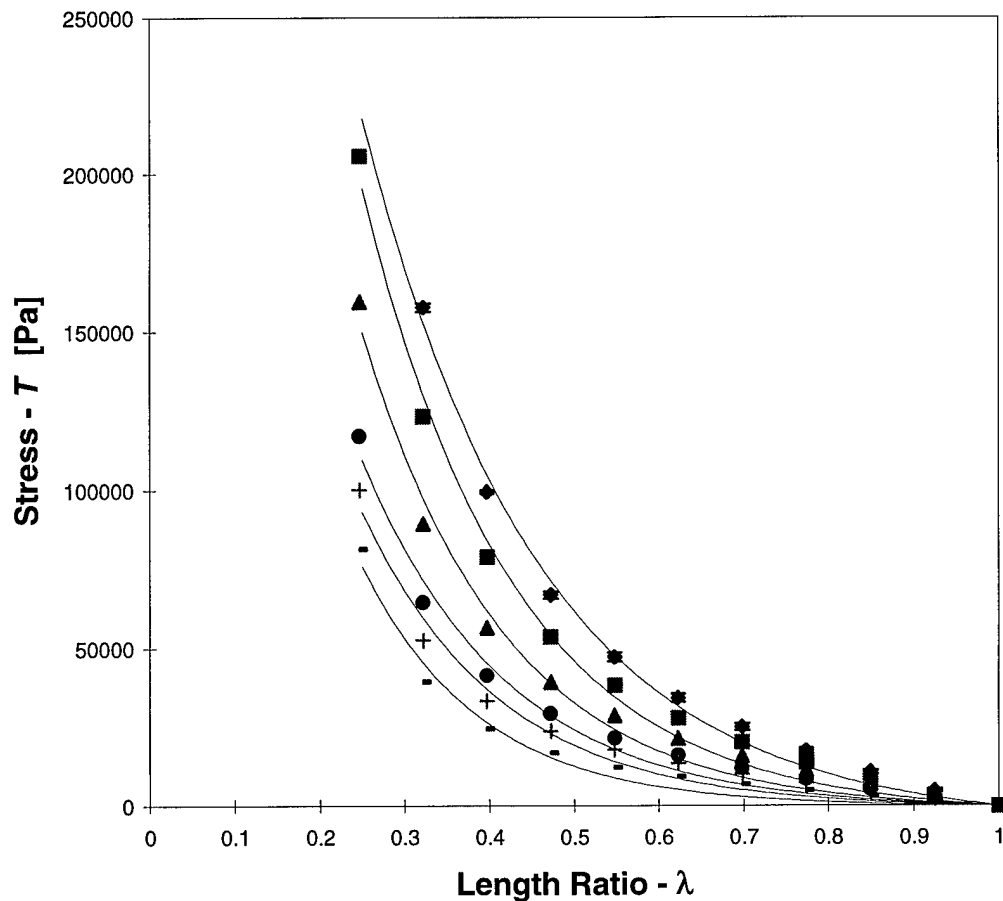


Figure 8: Uniaxial compression stress (Standard Method) as a function of length ratio and the corresponding exponential curve fit for the 6 Silicone materials used for the subjective test.

3.2 Subjective Experiment

The results of the subjective experiment are summarized in Fig. 9. The two-way ANOVA statistical test showed a significant difference between the performance obtained by the three tools ($p = 1.7 \cdot 10^{-6}$). The best performance in ranking the materials according to their stiffness was achieved by using the hand ($MSE=0.25$), whereas the worst performance was obtained by the standard endoscopic grasper ($MSE=3.15$). The performance of the FREG ($MSE=1.07$) was between the previous two, and closer to the performance of the hand than the grasper. The analysis suggests that there is no significant difference between the two operator groups (Surgeons and Control) that were tested ($p = 0.065$). From the results above, the null hypothesis (H_0) may be rejected (section 2.2.1). There is a significant difference between the performances of the three tools in ranking the materials according to their stiffness.

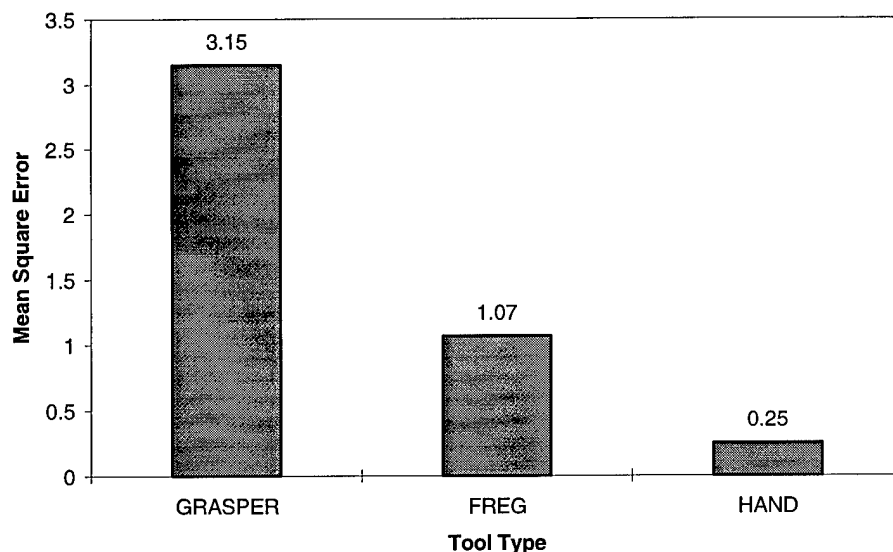


Figure 9: The Mean Square Error (MSE) of ranking the stiffness of 6 materials with three different tools.

4. Conclusions

Part of the haptic information that is lost when a surgeon manipulates a soft tissue using an endoscopic tool/grasper may be regained by using the bilateral force feedback technology implemented in endoscopic instrument. The FREG is capable of controlling the force or displacement of the jaw (tool tip) with interchangeable tools. To minimize cost and complexity, the system works with existing interchangeable reusable tools. The FREG controller was designed to maximize position control gain while preserving stability under unloaded conditions. Separating the human interface (finger loops) from the endoscopic tool allows one to generate a new Human-Machine interface (transfer function) in a way which enhances performance by overcoming the distortion which exists in the current mechanical endoscopic grasper setup.

The first phase in this two phase study was the objective experiment. The scatter plot (Fig 7) shows that the FREG in automatic mode is capable of discriminating between different soft tissues and latex/Silicone simulated tissue, and that the material's intrinsic biomechanical parameters can be identified for compression conditions. Moreover, a correlation from the mechanical characteristic perspective between the latex material and the soft tissue was found. For example, the data indicate that the latex material L2 might simulate the Liver. The α and β parameter values may be used to design this material for simulation purposes. The α parameter values for hollow organs (e.g. Colon, Small Bowel, Stomach) tend to be lower than the α parameter values for solid organs (e.g. Spleen, Liver). Although hollow organs have greater macroscopic variability (mucosa, muscular layers, and serosa) than solid organs, they also contain luminal air which could explain their softer characteristics.

Different values of α and β parameters for the 6 Silicone materials were obtained using the FREG compared to a testing method in which the contact areas were parallel. This result suggests that the measurements performed by the FREG are tool dependent. However, the two methods show the same trends. For example, the product $\alpha \cdot \beta$ may be an indicator for the material stiffness. The current values of the α and β parameters may be used for designing materials as tissue replication for training usage. However, since the material parameters measured by the FREG were found to be tool dependent, the usage of the α and β parameter values should be restricted to developing phantom materials for the purpose of training in MIS applications.

The second phase of this study focused on the psychophysics aspects of operating the FREG. The statistical analysis (2D ANOVA) of performance (*MSE*) measured for the 10 test operators ranking 6 materials according to their stiffness suggests significant improvement in the performance of the FREG relative to a standard endoscopic grasper. The FREG performance was closer to the human hand, in rating materials stiffness, which defines the upper performance limit, than the standard endoscopic grasper which defines the lower limit. Even in the hand in glove conditions, the test operators were not capable of ranking the material stiffness correctly in all the cases. This fact may raise the need for more advanced instruments like the FREG capable of increasing the haptic sensation beyond the capability of unaided hand.

The approach outlined in this study might be replicated in future studies with different endoscopic tools and control algorithms. This future research may study the soft tissue biomechanics under shear-compression conditions, and from another perspective, the correlation between the control algorithms and their parameters and the FREG performance. These studies may expand the current knowledge of the tissue/tool interface and on the human/machine interface in minimal invasive surgery.

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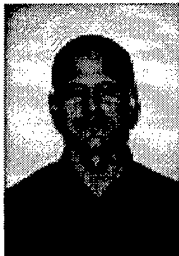
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Authors Biography



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Blake Hannaford received the B.S. degree in Engineering and Applied Science from Yale University in 1977, and the M.S. and Ph.D. degrees in Electrical Engineering from the University of California, Berkeley, in 1982 and 1985 respectively. Before graduate study, he held engineering positions in digital hardware and software design, office automation, and medical image processing. At Berkeley he pursued thesis research in multiple target tracking in medical images and the control of time-optimal voluntary human movement. From 1986 to 1989 he worked on the remote control of robot manipulators in the Man-Machine Systems Group in the Automated Systems Section of the NASA Jet Propulsion Laboratory, Caltech. He supervised that group from 1988 to 1989. Since September 1989, he has been at the University of Washington in Seattle, where he has been Associate Professor of Electrical Engineering since 1993. He was awarded the National Science Foundation's Presidential Young Investigator Award and the Early Career Achievement Award from the IEEE Engineering in Medicine and Biology Society. His currently active interests include haptic displays on the internet, surgical biomechanics, and biologically based design of robot manipulators. His lab URL is <http://rcs.ee.washington.edu/BRL>.



Mark P. MacFarlane received his B.A. in Biological Sciences from the California State University, Fullerton, and M.D. from the University of California, Davis. He completed general surgery residency at the University of California, San Francisco in 1997. His research experience includes a surgical oncology fellowship at the National Cancer Institute, National Institutes of Health in Bethesda, Maryland, which consisted of both basic and clinical research working on several large prospective clinical trials dealing with cytokines as a mode of therapy from advanced cancer. From 1997 to 1998 he had a fellowship in Minimal Invasive Surgery at the University of Washington, Seattle. His research was focused on clinical and basic research in the area of videoendoscopic surgery with emphasis on force feedback technology applications for telepresence and simulation. His current position is in private practice in general surgery, Spokane, Washington, with emphasis on surgical oncology and minimally invasive surgery.



Mika N. Sinanan received his M.D. from Johns Hopkins University in 1980, and a Ph.D. in Gastrointestinal Physiology from the University of British Columbia in 1991. He completed training in General and Gastrointestinal Surgery at the University of Washington in 1988 and joined the faculty the UW. Currently, he is an Associate Professor of Surgery and Adjunct Associate Professor of Electrical Engineering at the University of Washington. His research interests focus on technical surgical education, clinical application of advanced videoendoscopic Surgical procedures, and the study of force application and development of new instruments for technical maneuvers in minimally invasive surgery. He is a member of the American College of Surgeons, Society for Surgery of the Alimentary Tract, Society of American Gastrointestinal and Endoscopic Surgeons, and other regional societies.

Surgeon-Tool Force/Torque Signatures - Evaluation of Surgical Skills in Minimally Invasive Surgery

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Abstract

The best method of training for laparoscopic surgical skills is controversial. Some advocate observation in the operating room, while others promote animal and simulated models or a combination of surgical related tasks. The mode of proficiency evaluation common to all of these methods has been subjective evaluation by a skilled surgeon. In order to define an objective means of evaluating performance, an instrumented laparoscopic grasper was developed measuring the force/torque at the surgeon hand/tool interface. The measured database demonstrated substantial differences between experienced and novice surgeon groups. Analyzing forces and torques combined with the state transition during surgical procedures allows an objective measurement of skill in MIS. Teaching the novice surgeon to limit excessive loads and improve movement efficiency during surgical procedures can potentially result in less injury to soft tissues and less wasted time during laparoscopic surgery. Moreover the force/torque database measured in this study may be used for developing realistic virtual reality simulators and optimization of medical robots performance.

1. Introduction

One of the more difficult tasks in surgical education is to teach the optimal application of instrument forces and torques necessary to conduct an operation. This is especially problematic in the field of minimally invasive surgery (MIS) where the teacher is one step removed from the actual conduct of the operation. The use of virtual reality models for teaching these complex surgical skills has been a long-term goal of numerous investigators [1,2,3]. Developing such a system holds promise for providing a less stressful learning environment for the surgical student while eliminating any risk to the patient. In the development of such a system, it is important to understand the various components that comprise a realistic and useful training system [4]. While other studies have focused on the tool-tip/tissue interaction [5,6,7], the current research is aimed at analyzing the human/tool interface in MIS.

2. Purpose

The goal of this study is to create new quantitative knowledge of the forces and torques applied by surgeons on their instruments during minimally invasive surgery. Statistical models of this database can be used to characterize surgical skills for training advanced laparoscopic surgical procedures. Two areas in which this database might be applied are (i) Virtual Reality (developing haptic devices for realistic force feedback VR simulations of MIS procedures), and (ii) minimally invasive surgical robotics (optimizing mechanisms and actuators).

3. Methods

3.1 Experimental System Setup

Two types of information were acquired while performing MIS on pigs: (i) force/torque data measured at the human/tool interface and (ii) visual information of the tool tip interacting with the tissues. The two sources of information were synchronized in time, and recorded simultaneously for off line analysis. Protocols for anesthetic management, euthanasia, and survival procedures were reviewed and approved by the Animal Care Committee of the University of Washington and the Animal Use Review Division of the U.S. Army Veterinary Corps.

The forces and torques at the interface between the surgeon's hand and the endoscopic grasper handle were measured by two sensors. The first sensor was a three axis force/torque sensor (ATI - Mini model) which was mounted into the outer tube (proximal end) of a standard reusable 10 mm endoscopic grasper (Storz) - Fig. 1a. The sensor was capable of measuring simultaneously three components of force (F_x, F_y, F_z) and three components of torque (T_x, T_y, T_z) in the Cartesian frame (Fig. 1b). The forces and torques developed at the sensor location were a result of interacting with three interfaces in the

MIS environment: (i) hand-tool (ii) grasper tool-tip/tissue, and (iii) grasper outer-tube/trocar (port) in addition to the gravity and inertial loads. The summation of applied forces and torques were transferred through the grasper structure to the surgeon's hand and vice versa. The sensor orientation was such that X and Z axes generated a plane which was parallel to the tool's internal contact surface with the tissue in closing position, and the Y and Z axes defined a plane which was perpendicular to this surface (Fig.1b).

One of the grasper's mechanical features enabled the surgeon to rotate the grasper's outer tube, using a joint located near the handle, in order to change the orientation of the tool tip relative to the grasped tissue without changing the handle orientation. The alignment between the tool tip origin relative to the sensor remained unchanged since the outer tube and the tool tip were linked mechanically. A hole in the middle of the sensor allowed the rod inside the grasper tube to transfer the handle grasping force to the tool tip.

The second force sensor was mounted on the endoscopic grasper handle. Moving this handle caused the rod, sliding inside the outer tube, to transmit grasping/spreading forces from the surgeon's hand (F_g) to the tool tip. Due to this internal mechanism, whenever a grasping force was applied on the handle the outer tube was compressed. The outer tube compression was sensed by the force/torque sensor mounted within it. A nonzero force along the Z axis (F_z) would be developed during grasping, even if there were no external forces acting along this axis. This internal force coupling between the grasping/spreading and compression/tension along the Z axis was canceled by the processing software using a model of the grasper internal mechanism.

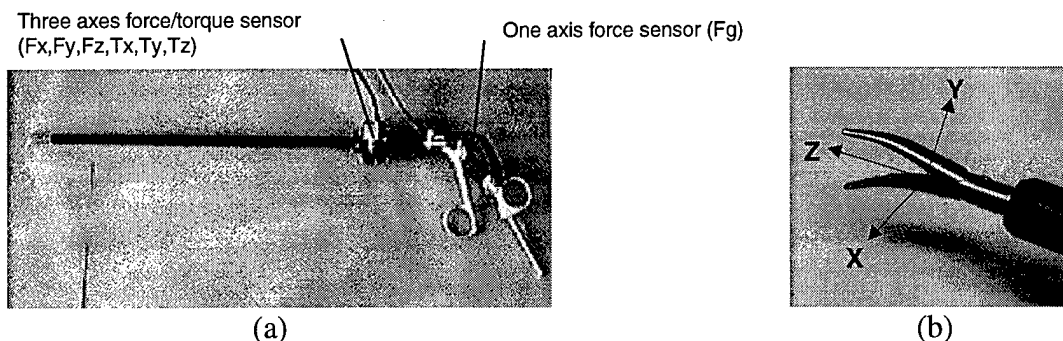


Figure 1: The instrumented endoscopic grasper: (a) The grasper with the three axis force/torque sensor implemented on the outer tube and a force sensor located on the instrument handle (b) The tool tip and X,Y,Z frame aligned with the three axis force/torque sensor.

The seven channels force/torque data were sampled at 30 Hz using a laptop with a PCMCIA 12 bit A/D card (National Instruments - DAQCard 1200). In addition, a LabView (National Instruments) application was developed incorporating a user interface for acquiring visualizing the force/torque data in real-time (Fig. 2).

The second source of information was the visual view from the endoscopic camera monitoring the movement of the grasper while interacting with the internal organs/tissues. This visual information was integrated with the force/torque human interface using a video mixer in a picture-in-picture mode and synchronized with time. The integrated interface was recorded during the surgical operation for off-line state analysis (Fig. 2).

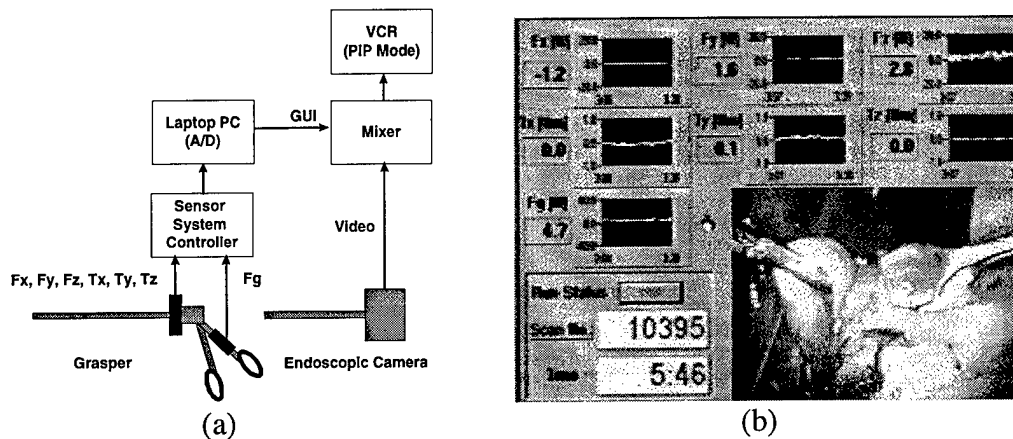


Figure 2: Experimental setup: (a) Block diagram of the experimental setup integrating the force/torque data and the view from the endoscopic camera, (b) Real-Time user interface of force/torque information synchronized with the endoscopic view of the procedure using picture-in-picture mode.

3.2 Surgical Experiment Setup and Clinical Trails

Four surgeons (two novice surgeons - NS and two experienced surgeons - ES) performed laparoscopic Cholecystectomy and laparoscopic Nissen Fundoplication in a porcine model (pig). Each operation was divided into steps (Table 1). Although all the steps were performed in each procedure, data were recorded only when the grasper was used with the following tool tips: atraumatic grasper, curved dissector, Babcock grasper.

3.3 Data Analysis

Two types of analysis were performed on the raw data: (i) video state analysis (SA) encoding the type of the tool-tip/tissue interaction into states and (ii) vector quantization (VQ) encoding the force/torque data into clusters (signatures). Each step of the operation was further divided into 17 different discrete tool maneuvers (states) in which the endoscopic tool was interacting with the tissue (Table 2). Each identified surgical maneuver (state), had a unique force/torque pattern. For example in the laparoscopic Cholecystectomy, isolation of the cystic duct and artery involves performing repeated pushing and spreading maneuvers which in turn requires to apply pushing forces mainly along the Z axis (F_z) and spreading forces (F_g) on the handle. Two expert surgeons independently performed frame by frame SA of the videotape with similar results.

Procedure	Step	Description	Tool Type	Hand	Video F/T
Laparoscopic Cholecystectomy	1	Positioning Gall Bladder	Atraumatic Grasper	L	+
	2	Exposure of Cystic Duct	Curved Dissector	R	+
	2*	Divide of Cystic Duct	Scissors	R	-
	3	Dissection of GallBladder Fossa	Curved Dissector	R	+
	4	Exposure of Cystic Artery	Curved Dissector	R	-
	4*	Dividing Artery	Scissors	R	-
Laparoscopic Nissen Fundoplication	1	Dissect Right Crus	Surgiwand	R	-
	2	Dissect Left Crus	Surgiwand	R	-
	3	Dissect Esophagus / Blunt	Curved Dissector	R	+
	4	Placing a Wrap Around the Esophagus	Babcock Grasper	R	+
	5	Suture Wrap / Intracorporeal Knot Tying With Needle Holder	Curved Dissector	R	+
	6	Coronal Sutures / Intracorporeal Knot Tying Endostitch	Endostitch	R	-

Table 1: Definitions of surgical procedure steps and types of the tool tip (Shaded steps performed but not recorded).

The 17 states can be divided into three types based on the number of movements performed simultaneously. The fundamental maneuvers were defined in type I. The idle state was defined as moving the tool in space without touching any internal organ, and the forces and torques developed in this state represented mainly the interaction with the trocar and the abdominal wall in addition to the gravitational and inertial forces. In the grasping and spreading states, compression and tension were being applied on the tissue by closing/opening the grasper handle. In the pushing state compression was applied on the tissue by moving the tool along the Z axis. For sweeping and lateral retraction, the tool was placed in one position while rotating it around the X and Y axes (trocar frame). The difference between sweeping and lateral retraction was that lateral retraction was a step-like movement as opposed to sweeping, which was a continuous movement. The rest of the states in groups II and III were combinations of the fundamental states of group I.

The second type of analysis used VQ algorithm to encode the multi dimensional force/torque data ($F_x, F_y, F_z, T_x, T_y, T_z, F_g$) into discrete symbols representing clusters (signatures). First the 7D force/torque data vector was reduced to a 5D vector by calculating the magnitude of the force and torque in the XY plane (F_{xy}, T_{xy}). Then The K-means algorithm was used to cluster the data into force/torque signatures of each one of the state defined in Table 2. Each force/torque signature represented a cluster center in a 5 dimensional space.

Type	State Name	State Acronym	Force / Torque						
			Fx	Fy	Fz	Tx	Ty	Tz	Fg
I	Idle	ID	*	*	*	*	*	*	*
	Grasping	GR							+
	Spreading	SP							-
	Pushing	PS			-				
	Sweeping	SW	+/-	+/-		+/-	+/-		
	Lateral Retraction	LR	+/-	+/-		+/-	+/-		
II	Grasping - Pulling	GR-PL			+				+
	Grasping - Pushing	GR-PS			-				+
	Grasping - Sweeping	GR-SW	+/-	+/-		+/-	+/-		+
	Grasping - Lateral Retraction	GR-LR	+/-	+/-		+/-	+/-		+
	Pushing - Spreading	PS-SP			-				-
	Pushing - Sweeping	PS-SW	+/-	+/-	-	+/-	+/-		
	Sweeping - Spreading	SW-SP	+/-	+/-		+/-	+/-		-
III	Grasping - Pulling - Sweeping	GR-PL-SW	+/-	+/-	+	+/-	+/-		+
	Grasping - Pushing - Sweeping	GR-PS-SW	+/-	+/-	-	+/-	+/-		+
	Pushing - Sweeping - Spreading	PS-SW-SP	+/-	+/-	-	+/-	+/-		-
	Pulling - Sweeping - Spreading	PL-SW-SP	+/-	+/-	+	+/-	+/-		-

Table 2: Definition of states and the corresponding directions of forces and torques applied in Cholecystectomy and Nissen fundoplication during MIS.

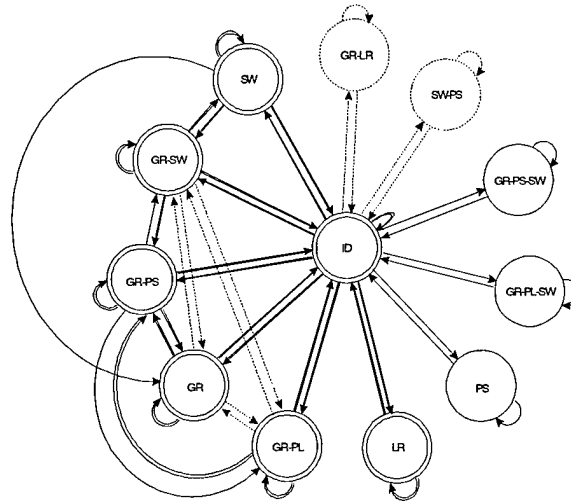
4. Results

A typical result of the state analysis was summarized for placing a wrap around the esophagus during laparoscopic Nissen fundoplication in Fig. 3. The state transition diagram had a shape of a star with a center point including the idle state (Fig 3a). This state was mainly used by both expert and novice surgeons to move from one operative state to the other. However the expert surgeons used the idle state only as a transition state while the novice spent significant amount of time in this state (Fig 3b). In general, it took to the novice surgeons 270% more time than the experienced surgeons to perform the same operation.

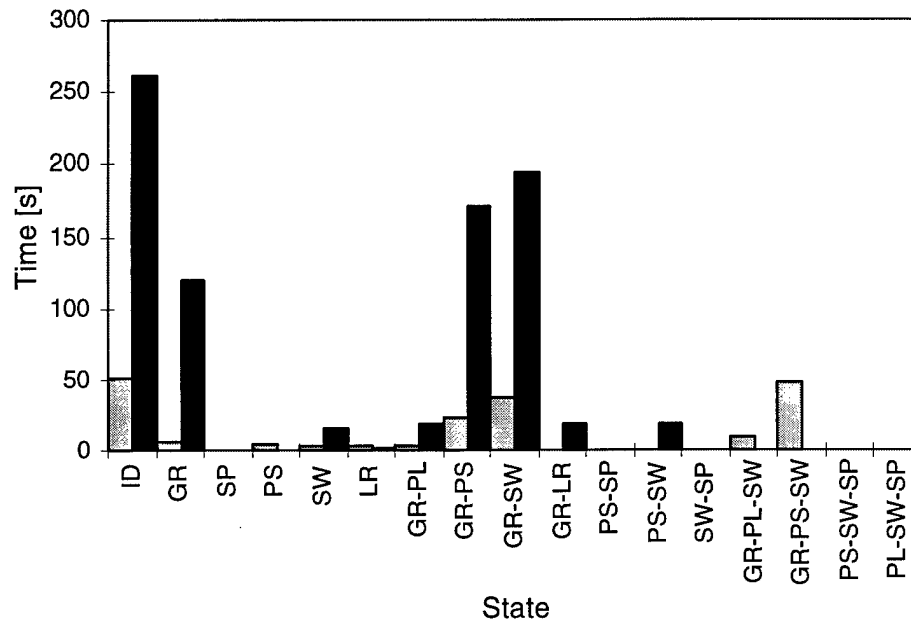
The force/torque data were plotted in a 3D space showing the loads developed at the sensor location while placing a wrap around the esophagus (laparoscopic Nissen fundoplication) - Fig. 4a. Using the state analysis and dividing each step of the operation into states, the force/torque segments for each state were lumped together. Figure 4b shows the force/torque distribution of the grasping-pulling state with respect to the normalized time spent in this state. Using the VQ algorithm the force/torque data of each state were further divided into clusters (Signatures). Figure 5 shows three typical signatures of the grasping-pulling state. The forces F_z and F_g were dominant in this signatures, whereas the rest of the force/torque values remained relatively low. This three clusters may represent the entire force/torque space of the grasping-pulling state, and the rest of the data in these state can be correlated with this three signatures. Analyzing the

data of the experienced and novice surgeons showed that the forces and torques used to perform an operation was 130%-138% greater for novice surgeons

(a)



(b)

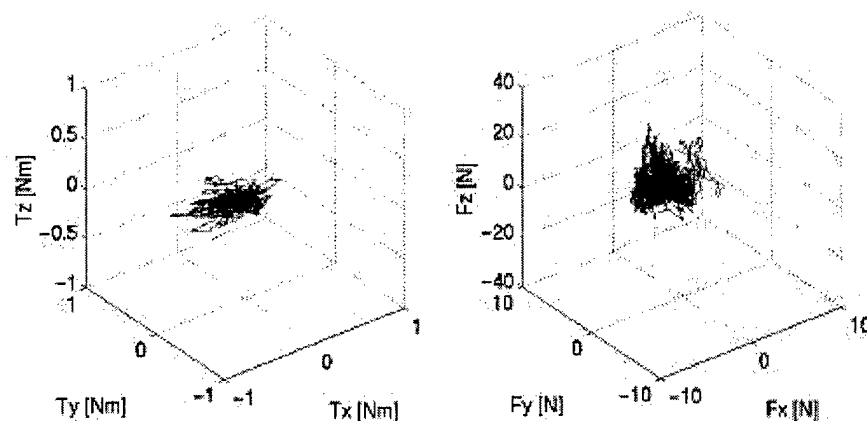


(a)

(b)

Figure 3: State analysis of placing wrap around the esophagus during laparoscopic Nissen fundoplication: (a) State transitions (solid line - expert surgeon, dashed line - novice surgeon, doubled line - both) (b) Time sharing between states (■ - Experienced Surgeon, ■ - Novice Surgeon)

(a)



(b)

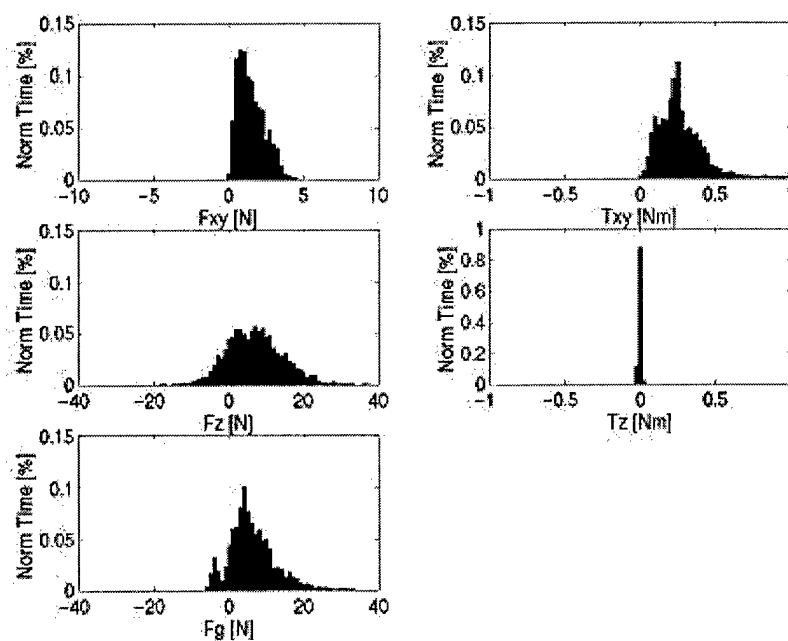


Figure 4: Force/torque data measured during placing wrap around esophagus (laparoscopic Nissen fundoplication) by an experienced surgeon: (a) Raw force/torque data, (b) Force/torque data distribution during grasping-pulling state with respect to the normalized time.

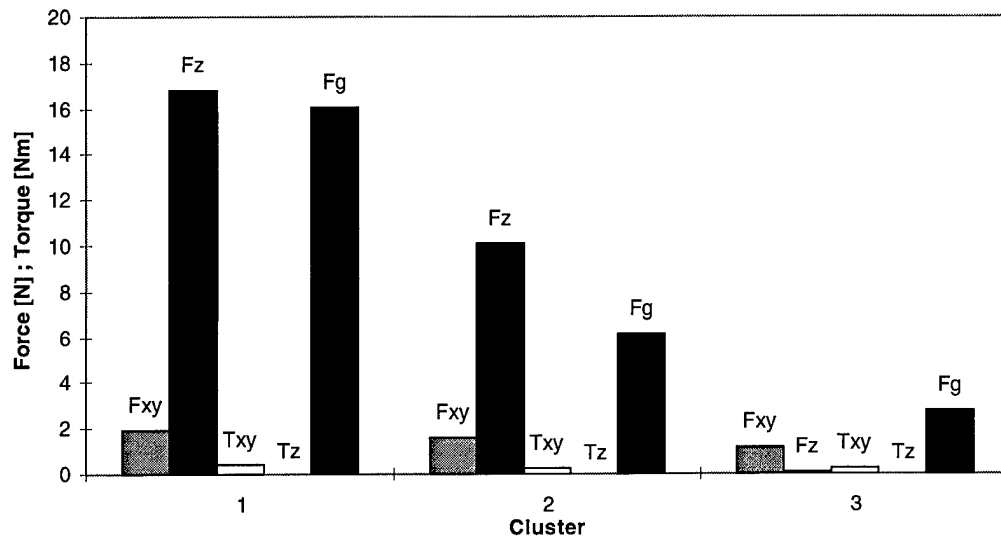


Figure 5: Force/torque signatures of the grasping-pulling state

5. Conclusions

Minimally invasive surgery is a complex task which requires a synthesis between visual and haptic information. Analyzing MIS in terms of these two sources of information is a key step towards developing objective criteria for training surgeons and evaluating the performance of a master/slave robotic system for teleoperation or a haptic device for virtual reality simulations. The state transition data and the force/torque signatures are objective criteria for evaluating skills and performance in MIS. In general, it took the expert surgeon less time while applying less forces and torques to perform a typical MIS compared to the novice surgeon. This may be a result of advanced knowledge of the anatomy, higher level of eye-hand coordination and greater experience in handling the endoscopic surgical instrument.

The approach outlined in this study could be extended by increasing the size of the database which will allow development of statistical models like the Hidden Markov Model (HMM) of surgical procedures. This information, combined with other feedback data, may be used as a basis to develop teaching techniques for optimizing tool usage in MIS. The novice surgeons could practice these skills outside of the operating room on animal models or by using realistic virtual reality simulators, until they had achieved the desired level of competence, and compare themselves to norms established by experienced surgeons.

Acknowledgments

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Skills Evaluation in Minimally Invasive Surgery Using Force/Torque Signatures

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Abstract

Background: One of the more difficult tasks in surgical education is to teach the optimal application of instrument forces and torques necessary to facilitate the conduct of an operation. For laparoscopic surgery, such training has traditionally taken place in the operating room, reducing operating room efficiency and potentially affecting the safe conduct of the operation. **Objective:** The objective of the current study was to measure and compare forces and torque signatures at the tool/hand interface generated during laparoscopic surgery by novice (NS) and experienced (ES) surgeons using a novel force-torque laparoscopic instrument. **Methods:** Four surgeons (2-NS, 2-ES) performed a cholecystectomy and Nissen fundoplication in a porcine model. An instrumented laparoscopic grasper with interchangeable standard surgical tips equipped with a three-axis force/torque sensor was used to measure the force/torque signature at the hand/tool interface. Force/torque data synchronized with visual view of the tool operative maneuvers were collected simultaneously via a novel graphic user interface incorporated picture-in-picture video technology. Subsequent frame by frame, video analysis of the operation allowed to define states within each step of the operation. Forces and torques measured within each state were further analyzed using vector quantization and the hidden Markov statistical model. **Results:** The NS group used a mean of 138% greater force and 130% greater torque during all stages of an operation compared to the ES group. Furthermore, the completion time of a laparoscopic cholecystectomy on a porcine model was 270% greater in the NS group. State analysis suggests that the majority of this time was consumed in an "idle" without useful contribution to the conduct of the operation. **Conclusions:** Preliminary data suggest that force/torque signatures provide an objective means of distinguishing novice from skilled surgeons. Clinical force/torque signature analysis using the proposed technology and methodology may be helpful in both training and measuring technical proficiency during laparoscopic surgery.

Keywords: laparoscopy, training, technical skills, force torque signatures

1. Introduction

One of the more difficult tasks in surgical education is to teach the optimal application of instrument handling necessary to conduct an operation. This is especially problematic in the field of minimally invasive surgery (MIS) where the teacher is one step removed from actual tissue contact. For laparoscopic surgery, such training has traditionally taken place in the operating room; thereby reducing operating room efficiency and potentially affecting the safe conduct of the operation.

The use of virtual reality models for teaching these complex surgical skills has been a long-term goal of numerous investigators [1,2,3]. Developing such a system holds promise for providing a less stressful learning environment for the surgical novice while eliminating any risk to the patient. Development of such a system requires an understanding of the various components that comprise a realistic and useful training system [4]. While other studies have focused on the tool-tip/tissue interaction and deformation [5, 6, 7], this research measures the forces and torques applied at the human/tool interface in MIS.

The objective of this study is to measure and compare forces and torque signatures generated during laparoscopic surgery by novice (NS) and experienced (ES) surgeons using a novel instrumented endoscopic grasper. Statistical models of the quantitative knowledge gained can be used to characterize surgical skills for training novice surgeons in the performance of laparoscopic procedures. Two areas in which the force/torque signature database might be used are (1) virtual reality (developing haptic devices for realistic force feedback VR simulations of MIS procedures), and (2) minimally invasive surgical robotics (optimizing mechanisms and actuators).

2. Materials and Methods

2.1 Subjects and Protocol

Four surgeons (two novice-NS, and two experienced -ES) performed a laparoscopic cholecystectomy and laparoscopic Nissen fundoplication in a porcine model. Protocols for anesthetic management, euthanasia, and survival procedures were reviewed and approved by the Animal Care Committee of the University of Washington and the Animal Use Review Division of the U.S. Army Veterinary Corps. Each operation was subdivided into steps (Table 1). Although all the steps were performed in each procedure, data were recorded only when the instrumented endoscopic tool was used with the following tool tips: atraumatic grasper, curved dissector, or Babcock grasper.

2.2 Experimental System Setup

Two types of information were acquired while performing laparoscopic procedures (1) force/torque data measured at the human/tool interface (2) visual information of the tool tip interacting with the tissues. The two sources of information were synchronized in time, and recorded simultaneously for subsequent analysis using picture in picture video technology.

The forces and torques at the interface between the surgeons' hand and the endoscopic grasper handle were measured by two sensors. The first sensor was a three axis force/torque sensor (ATI-Mini model) which was mounted into the outer tube (proximal end) of a standard reusable 10 mm endoscopic grasper (Storz) - Fig. 1a. The sensor was capable of simultaneous measurements of three components of force (F_x , F_y , F_z) and three components of torque (T_x , T_y , T_z) in a Cartesian frame (Fig. 1b). The summation of applied forces and torques were transferred through the grasper structure to the surgeon's hand, as occurs in a normal instrument, and vice versa. The sensor orientation was such that X and Z axes formed a plane parallel to the tool's internal jaw contact surface with the jaw closed and the Y and Z axes defined a plane

perpendicular to that surface (Fig. 1b). A second force sensor was mounted to the endoscopic grasper handle to permit the measurement of grasping force applied by the surgeon on the instrument.

The grasper had a reticulating feature which enabled the surgeon to change the orientation of the tool tip relative to the grasped tissue without changing the handle orientation. The alignment between tool tip origin relative to the sensor remained unchanged since the outer tube and the tool tip were linked mechanically.

Force/torque data were integrated with the laparoscopic camera view of instrument activity. Force and torque data were sampled at 30 Hz over seven channels using a laptop computer with a PCMCIA 12 bit A/D card (National Instruments - DAQCard 1200). A LabView (National Instruments) application was developed with a novel user interface for acquiring and visualizing the force/torque data in real-time (Fig.2) during an actual operation. The visual view from the endoscopic camera which monitored the movement of the grasper's tip while interacting with the internal organs/tissues. The visual information was integrated with the force/torque human interface using a video mixer in a picture-in-picture mode to permit correlation of force/torque data to instrument activity. The integrated interface was recorded during the operation for subsequent frame by frame analysis (Fig. 2).

2.3 Data analysis

Two types of analysis were performed on the raw data: (1) visual analysis defining the type of tool-tip/tissue interaction - states and (2) vector quantization (VQ) encoding the force/torque data into clusters - signatures. Each step of the operation was analyzed and a total of 17 different discrete tool-tissue interactions or states were identified (Table 2). Upon further analysis, it became clear that each identified surgical maneuver (state) had a unique force/torque pattern. For example, isolation of the cystic duct and artery in laparoscopic cholecystectomy involves performing repeated pushing and spreading maneuvers which in turn require application of pushing forces mainly along the Z axis (F_z) and spreading forces (F_g)

on the handle. Two expert surgeons independently performed frame by frame state analysis of the videotape with similar results. Although discrete, the 17 states can be grouped into three broad types based on the number of movements performed simultaneously. Fundamental maneuvers were defined as Type 1, and included the idle state (moving the tool in space without touching any structures within the insufflated abdomen). The forces and torques used in this state represent mainly the interaction of the trocar with the abdominal wall in addition to gravitational and inertial forces. In the grasping and spreading states, compression and tension were applied to tissue by closing/opening the grasper handle. In the pushing state, compression was applied on the tissue by moving the tool along the Z axis. For sweeping, the tool was placed in one position while rotating around the X and Y axes (trocar frame). Type II and Type III were states defined as combinations of two or three states from group I (Table 2).

Force/torque analysis to define force/torque signatures used a VQ algorithm to encode the multi-dimensional force/torque data (F_x , F_y , F_z , T_x , T_y , T_z , F_g) into discrete symbols representing clusters. First the 7D force/torque data vector was reduced to a 5D vector by calculating the magnitude of the forces and torques in the XY plane (F_{xy} , T_{xy}). A Pattern recognition analysis (clustering analysis) known as the K-means algorithm was then used to cluster the data into force/torque signatures for each state defined in Table 2. Each force/torque signature represented a cluster centered in a 5 dimensional plane.

3. Results

Each step of an operation can be interpreted as a series of states and transitions between states to achieve the goal of a particular step of an operation. A typical result of state analysis was summarized in the step of placing a fundal wrap around the esophagus during the performance of a laparoscopic Nissen fundoplication (Fig. 3). For this maneuver, the state transition diagram had the shape of a star with the center point being the idle state (Fig 3a). Each circle represented a different state and the arrows stood for transitions between states.

This analysis clearly demonstrated several phenomena. First, expert and novice surgeons took different paths to reach the same goal (Figure 3a). Each group utilized states and transitions not used by the other group. For example, in placing the fundal wrap, the novice surgeon used the two states of sweeping/pushing, grasping/lateral retraction while the expert did not. Secondly, the expert surgeons used the idle state only as a transition state while the novice spent a significant amount of time in this state (Figure 3b). In general, it took the novice surgeons 270% more time than the experienced surgeons to perform the same operation.

The force/torque data were plotted in a 3D space showing the loads developed at the sensor location while placing a wrap around the esophagus in a laparoscopic Nissen fundoplication (Fig. 4). Using state analysis and dividing each step of the operation into states, the force/torque segments for each state were lumped together. Figure 4b. showed the force/torque segments for each state. Figure 5 showed the force/torque distribution of the grasping-pulling state with respect to the normalized time spent in this state. Using the VQ algorithm the force/torque data of each state were further divided into clusters (signatures). Figure 6 showed three typical signatures of the grasping-pulling state. The forces F_z and F_g were dominant in this state, whereas the rest of the force/torque values remained relatively low. The three clusters in Figure 6 represent the entire force/torque measurements within the grasping-pulling state. Data pooled from the entire data set show that the forces and torques used to perform an operation were 130-138% greater for novice surgeons.

4. Conclusions

Minimally invasive surgery is a complex task which requires a synthesis between visual and haptic (sense of touch) information. Analyzing MIS in terms of these two sources of information is a key step towards developing objective criteria for training surgeons. Distinct differences can be identified between novice and expert surgeons. These differences include the amount of time spent performing an operation (which should decrease with instrument and anatomic familiarity), as well as states used to conduct an operation

and transitions from state to state. The application of force and torque to perform each step of an operation also differs between groups. Novice surgeons use greater forces and torques than expert surgeons to perform the same operation. Ultimately, the information obtained from a system such as the one described in this manuscript could be used to develop a surgical training simulator. The novice surgeon could train outside the operating room until the expert level of performance is achieved.

Acknowledgments

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Legends

- Figure 1:** The instrumented endoscopic grasper: (a) The grasper with the three axis force/torque sensor implemented on the outer tube and a force sensor located on the instrument handle (b) The tool tip and x,y,z frame aligned with the three axis force/torque sensor.
- Figure 2:** Experimental setup: (a) Block diagram of the experimental setup integrating the force/torque data and the view from the endoscopic camera, (b) Real-time graphic user interface (GUI) of force/torque information synchronized with the endoscopic view of the procedure using picture-in-picture mode.
- Figure 3:** State analysis of placing wrap around the esophagus during laparoscopic Nissen fundoplication: (a) State transitions (solid line-expert surgeon, dashed line-novice surgeon, double line-both) (b) Time sharing between states (gray-experienced, black-novice).
- Figure 4:** Force/torque data measured while placing wrap around esophagus comparing expert (a) novice (b) surgeon.
- Figure 5:** Force/torque data distribution measured during placing wrap around esophagus
- Figure 6:** Force/torque signatures of the grasping-pulling state
- Table 1:** Definitions of surgical procedure steps and type of tool tip used (Shaded steps performed but not recorded).
- Table 2:** Definition of states and the corresponding directions of forces and torques applied in laparoscopic cholecystectomy and Nissen fundoplication.

Figure 1 a

Figure 1 b

Figure 2 a

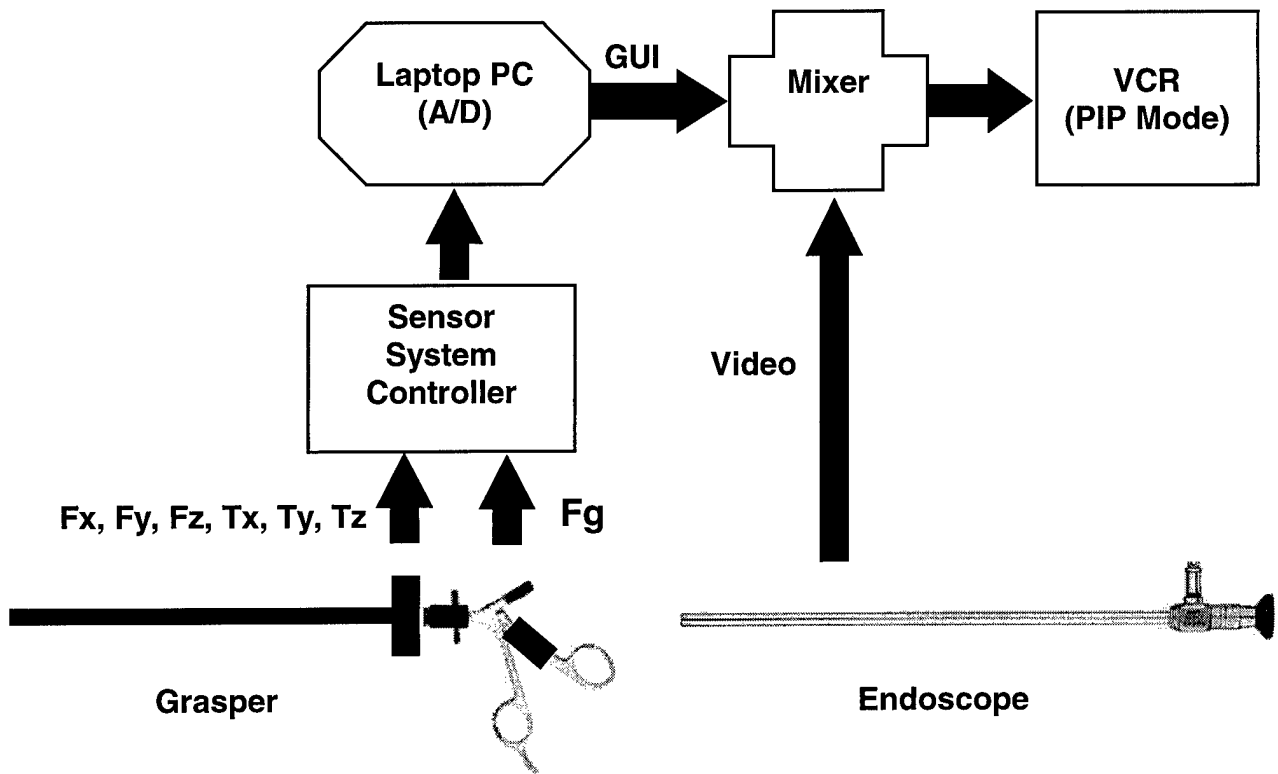


Figure 2b

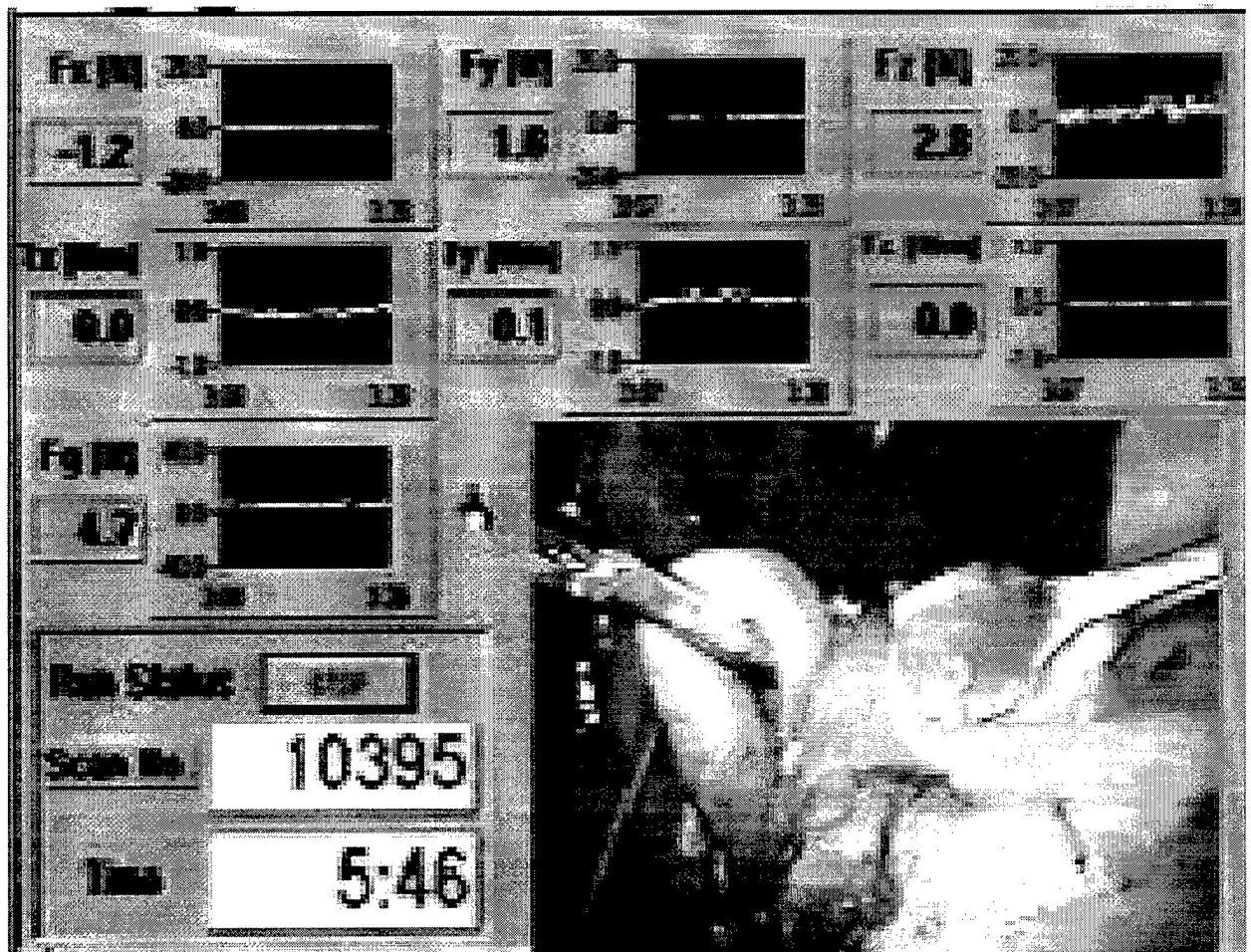


Figure 3a

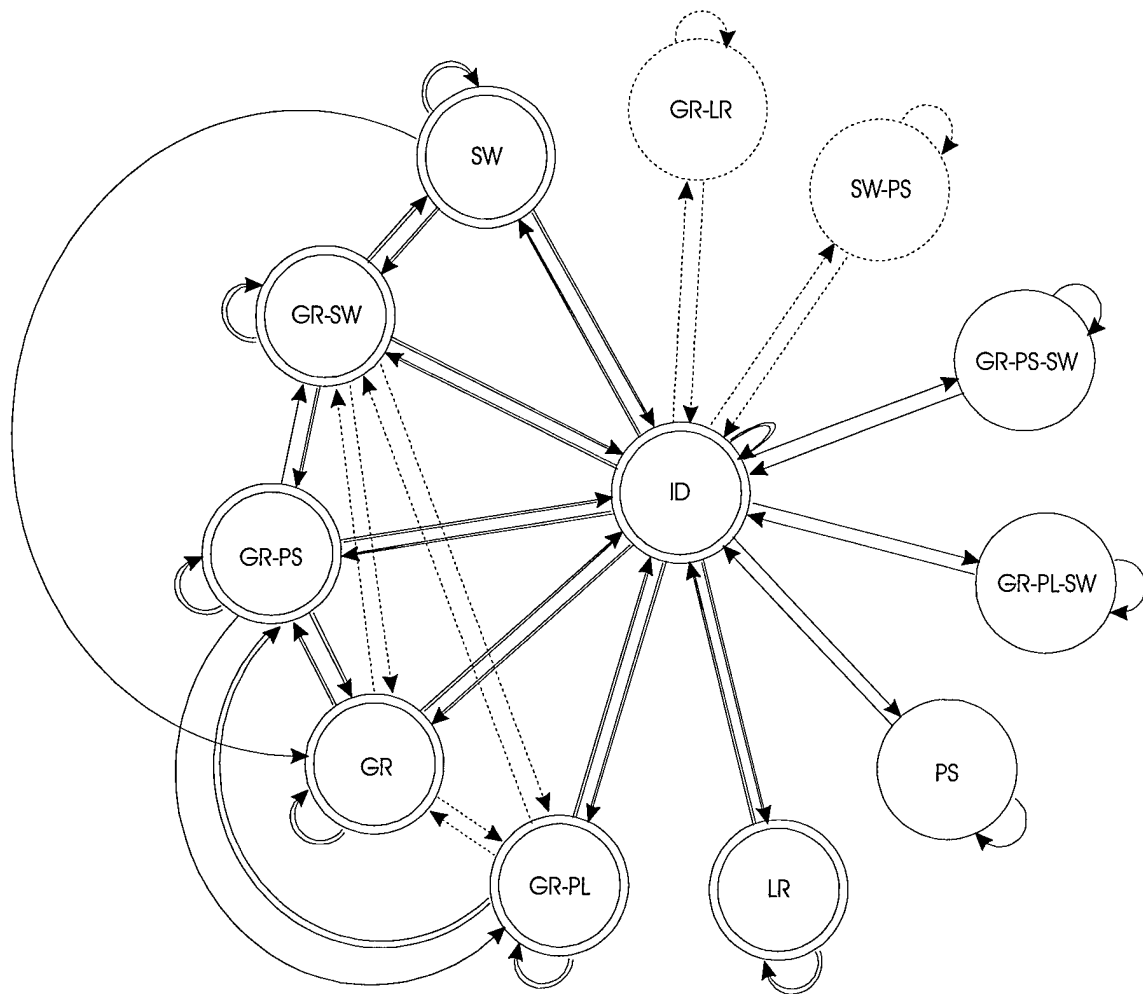


Figure 3b

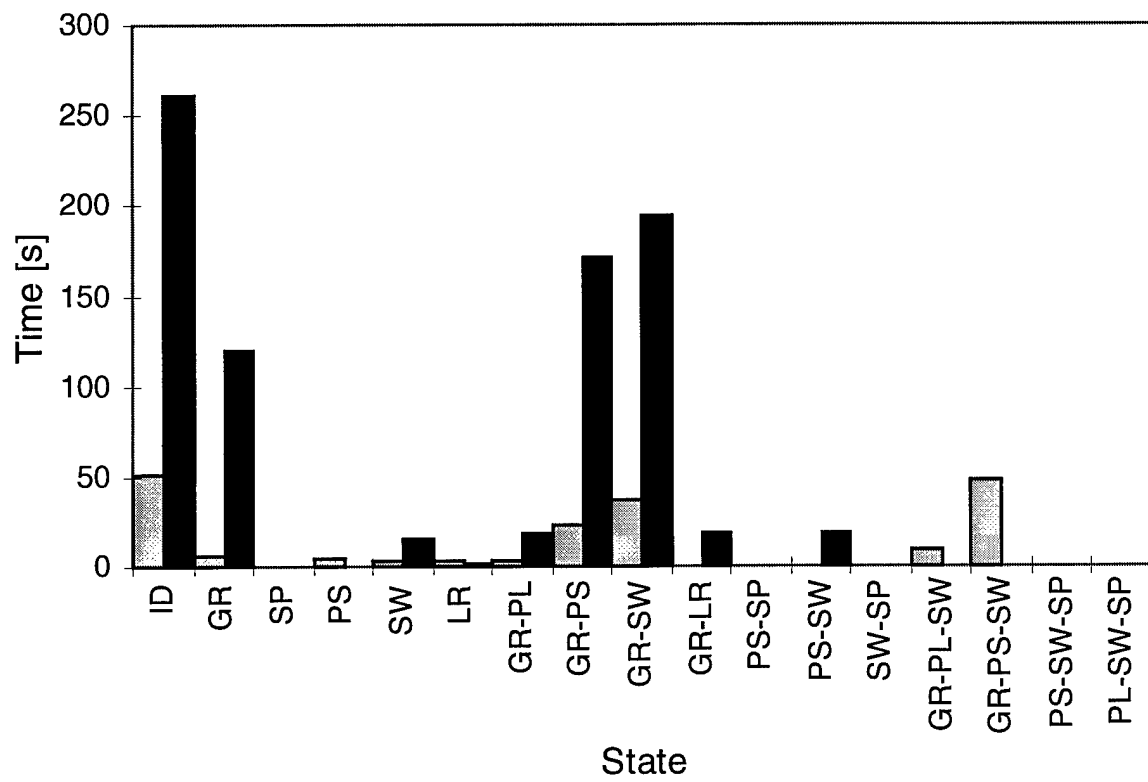


Figure 4a

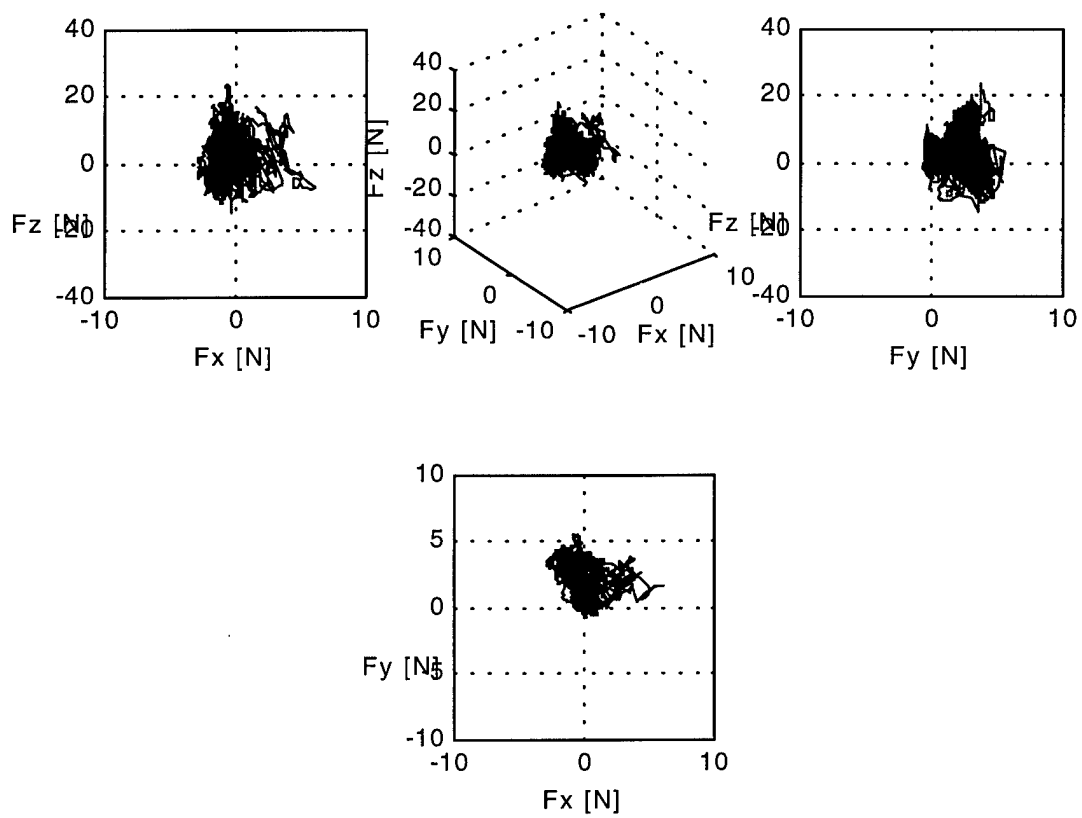


Figure 4b

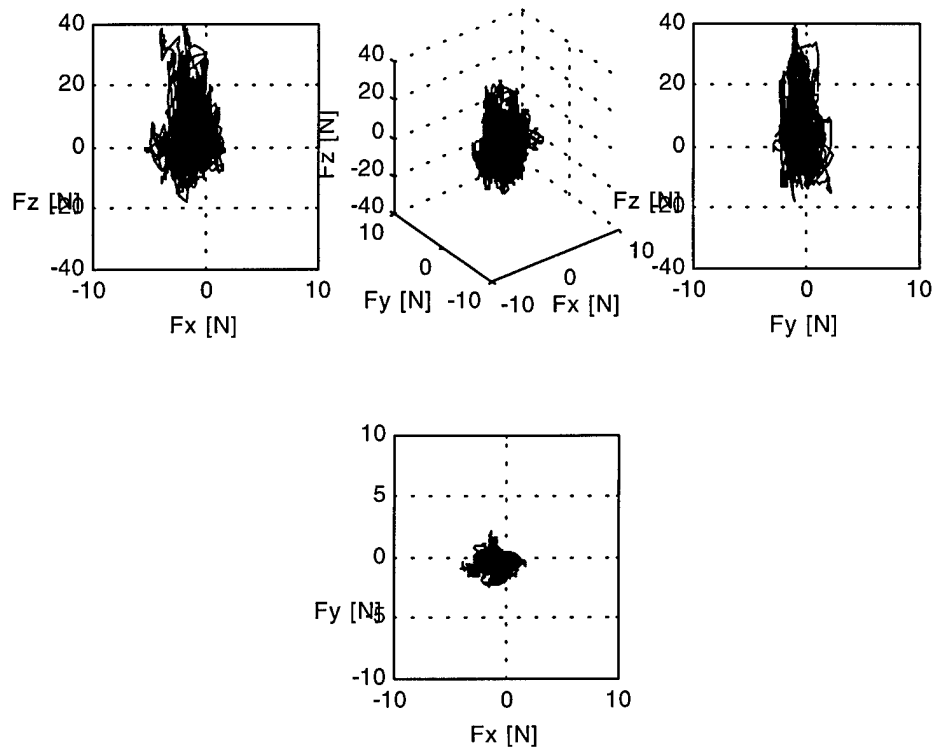


Figure 5

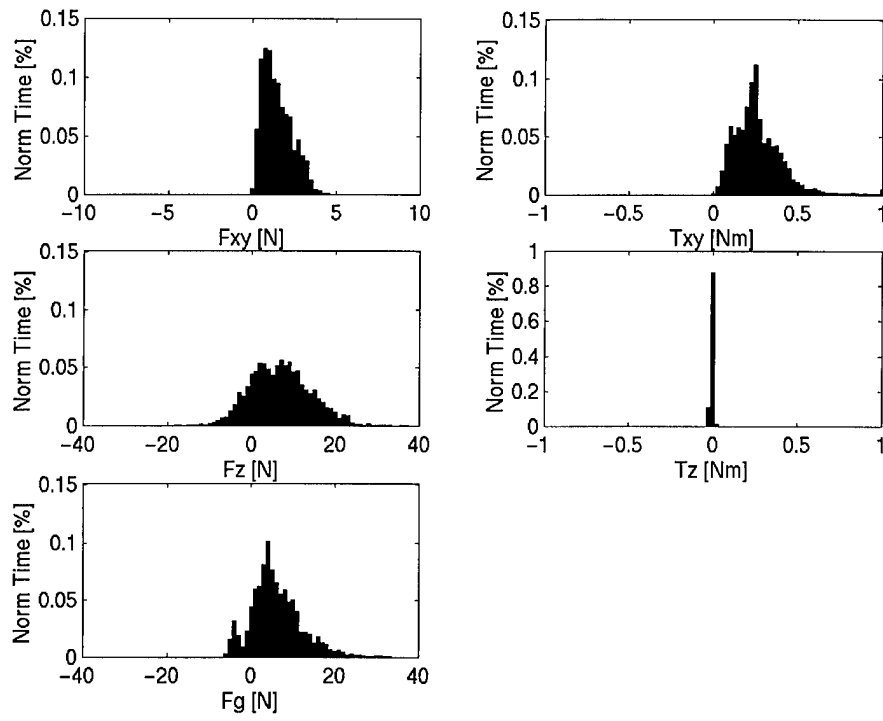


Figure 6

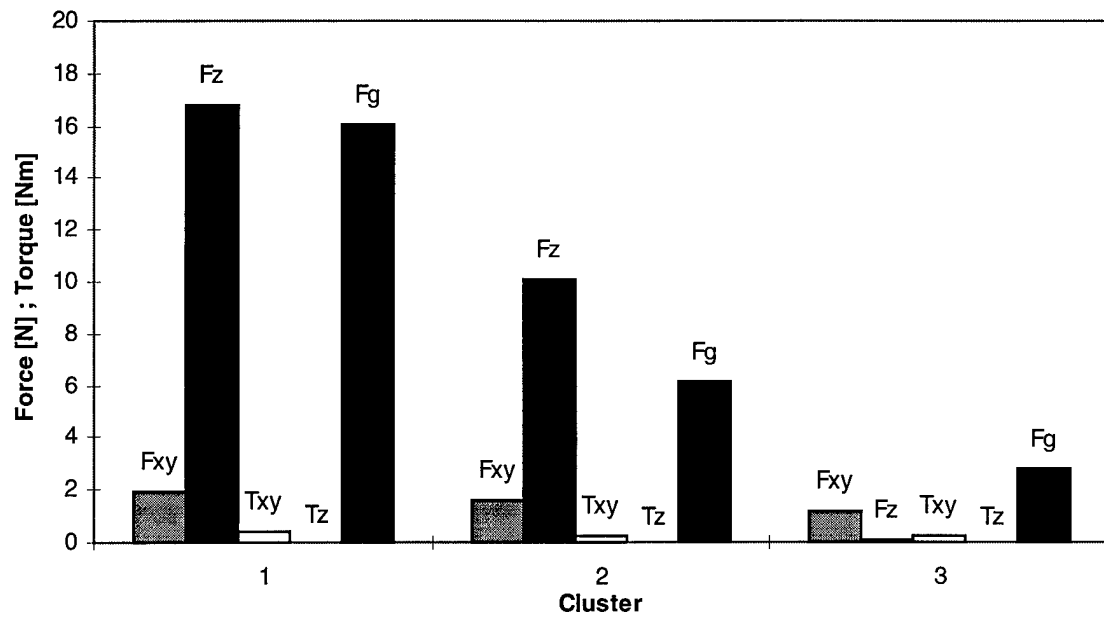


Table 1

Procedure	Step	Description	Tool Type	Hand	Video F/T
Laparoscopic Cholecystectomy	1	Positioning Gall Bladder	Atraumatic Grasper	L	+
	2	Exposure of Cystic Duct	Curved Dissector	R	+
	2*	Divide of Cystic Duct	Scissors	R	-
	3	Dissection of Gall Bladder Fossa	Curved Dissector	R	+
	4	Exposure of Cystic Artery	Curved Dissector	R	-
	4*	Division of Artery	Scissors	R	-
Laparoscopic Nissen Fundoplication	1	Dissect Right Crus	Surgiwand	R	-
	2	Dissect Left Crus	Surgiwand	R	-
	3	Dissect Esophagus / Blunt	Curved Dissector	R	+
	4	Placing a Wrap Around the Esophagus	Babcock Grasper	R	+
	5	Suture Wrap / Intracorporeal Knot Tying With Needle Holder	Curved Dissector	R	+
	6	Coronal Sutures / Intracorporeal Knot Tying Endostitch	Endostitch	R	-

Table 2

Type	State Name	State Acronym	Force / Torque						
			Fx	Fy	Fz	Tx	Ty	Tz	Fg
<i>I</i>	Idle	ID	*	*	*	*	*	*	*
	Grasping	GR							+
	Spreading	SP							-
	Pushing	PS			-				
	Sweeping	SW	+/-	+/-		+/-	+/-		
	Lateral Retraction	LR	+/-	+/-		+/-	+/-		
<i>II</i>	Grasping - Pulling	GR-PL			+				+
	Grasping - Pushing	GR-PS			-				+
	Grasping - Sweeping	GR-SW	+/-	+/-		+/-	+/-		+
	Grasping - Lateral Retraction	GR-LR	+/-	+/-		+/-	+/-		+
	Pushing - Spreading	PS-SP			-				-
	Pushing - Sweeping	PS-SW	+/-	+/-	-	+/-	+/-		
	Sweeping - Spreading	SW-SP	+/-	+/-		+/-	+/-		-
<i>III</i>	Grasping - Pulling - Sweeping	GR-PL-SW	+/-	+/-	+	+/-	+/-		+
	Grasping - Pushing - Sweeping	GR-PS-SW	+/-	+/-	-	+/-	+/-		+
	Pushing - Sweeping - Spreading	PS-SW-SP	+/-	+/-	-	+/-	+/-		-
	Pulling - Sweeping - Spreading	PL-SW-SP	+/-	+/-	+	+/-	+/-		-